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Term:

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L8 and ((identical$3 or "substantially the same"
or "the same" or equal or equivalent) with ("field
of view" or FOV or field-of-view))
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DATE: Friday, March 08, 2002 Printable Copy Create Case

Set Name side by side	**Query**	**Hit Count**	**Set Name** result set
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L9	L8 and ((identical$3 or "substantially the same" or "the same" or equal or equivalent) with ("field of view" or FOV or field-of-view))	4	L9
L8	L7 and ((second or third or another or additional or secondary or tertiary) with coil)	32	L8
L7	L6 and (coil with array)	42	L7
L6	L5 and (array)	152	L6
L5	L4 and (radio-frequency or "radio frequecy" or RF)	260	L5
L4	l3 and ((image or imaging or imaged) with ("field of view" or FOV or field-of-view))	1389	L4
L3	("field of view" or FOV or field-of-view)	3522	L3
L2	(Srinivasan with Ravi)[IN]	33	L2
L1	(Srinivasan)[IN]	1721	L1

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Search Results - Record(s) 1 through 32 of 32 returned.

☐ 1. Document ID: US 20020021128 A1

L8: Entry 1 of 32 File: PGPB Feb 21, 2002

PGPUB-DOCUMENT-NUMBER: 20020021128
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020021128 A1

TITLE: Magnetic resonance imaging involving movement of patient's couch

PUBLICATION-DATE: February 21, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Kuhara, Shigehide	Otawara-Shi		JP	

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 2. Document ID: US 20020013526 A1

L8: Entry 2 of 32 File: PGPB Jan 31, 2002

PGPUB-DOCUMENT-NUMBER: 20020013526
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020013526 A1

TITLE: Inherently de-coupled sandwiched solenoidal array coil

PUBLICATION-DATE: January 31, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Su, Sunyu	South San Francisco	CA	US	
Kaufman, Leon	San Francisco	CA	US	
Arakawa, Mitsuaki	Hillsborough	CA	US	

US-CL-CURRENT: 600/422; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 3. Document ID: US 20010043068 A1

L8: Entry 3 of 32 File: PGPB Nov 22, 2001

PGPUB-DOCUMENT-NUMBER: 20010043068
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20010043068 A1

TITLE: Method for parallel spatial encoded MRI and apparatus, systems and other methods related thereto

PUBLICATION-DATE: November 22, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Lee, Ray F.	Clifton-Park	NY	US	

US-CL-CURRENT: 324/309; 324/307, 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 4. Document ID: US 6326786 B1

L8: Entry 4 of 32 File: USPT Dec 4, 2001

US-PAT-NO: 6326786
DOCUMENT-IDENTIFIER: US 6326786 B1

TITLE: Magnetic resonance imaging method and apparatus

DATE-ISSUED: December 4, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Pruessmann; Klaas P.	Zurich			CHX
Weiger; Markus	Dietikon			CHX
Scheidegger; Markus B.	Bisikon			CHX
Boesiger; Peter	Ennetbaden			CHX

US-CL-CURRENT: 324/312; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 5. Document ID: US 6289232 B1

L8: Entry 5 of 32 File: USPT Sep 11, 2001

US-PAT-NO: 6289232
DOCUMENT-IDENTIFIER: US 6289232 B1

TITLE: Coil array autocalibration MR imaging

DATE-ISSUED: September 11, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jakob; Peter M.	Brookline Village	MA		
Sodickson; Daniel K.	Cambridge	MA		
Griswold; Mark	Brookline	MA		

US-CL-CURRENT: 600/410; 324/307, 324/309, 324/318, 324/322, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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6. Document ID: US 6278275 B1

L8: Entry 6 of 32 File: USPT Aug 21, 2001

US-PAT-NO: 6278275
DOCUMENT-IDENTIFIER: US 6278275 B1

TITLE: Gradient coil set with non-zero first gradient field vector derivative

DATE-ISSUED: August 21, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros S.	Solon	OH		
Schlitt; Heidi A.	Chesterland	OH		

US-CL-CURRENT: 324/318; 324/309, 324/320

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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7. Document ID: US 6215911 B1

L8: Entry 7 of 32 File: USPT Apr 10, 2001

US-PAT-NO: 6215911
DOCUMENT-IDENTIFIER: US 6215911 B1

TITLE: Method for correcting the intensity of an image

DATE-ISSUED: April 10, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Goertler; Georg	Baiersdorf			DEX
Schnur; Guenter	Hemhofen			DEX

US-CL-CURRENT: 382/264; 382/169, 382/260, 382/274

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 8. Document ID: US 6177797 B1

L8: Entry 8 of 32 File: USPT Jan 23, 2001

US-PAT-NO: 6177797
DOCUMENT-IDENTIFIER: US 6177797 B1

TITLE: Radio-frequency coil and method for resonance/imaging analysis

DATE-ISSUED: January 23, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 9. Document ID: US 6150816 A

L8: Entry 9 of 32 File: USPT Nov 21, 2000

US-PAT-NO: 6150816
DOCUMENT-IDENTIFIER: US 6150816 A

TITLE: Radio-frequency coil array for resonance analysis

DATE-ISSUED: November 21, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 10. Document ID: US 6104943 A

L8: Entry 10 of 32 File: USPT Aug 15, 2000

US-PAT-NO: 6104943
DOCUMENT-IDENTIFIER: US 6104943 A

TITLE: Phased array echoplanar imaging system for fMRI

DATE-ISSUED: August 15, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Frederick; Blaise deB.	Lexington	MA		
Wald; Lawrence	Cambridge	MA		
Renshaw; Perry F.	Arlington	MA		

US-CL-CURRENT: 600/410; 324/309, 324/318, 600/421

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 11. Document ID: US 6084411 A

L8: Entry 11 of 32 File: USPT Jul 4, 2000

US-PAT-NO: 6084411
DOCUMENT-IDENTIFIER: US 6084411 A

TITLE: Flexible lightweight attached phased-array (FLAP) receive coils

DATE-ISSUED: July 4, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Giaquinto; Randy Otto John	Burnt Hills	NY		
Dumoulin; Charles Lucian	Ballston Lake	NY		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 12. Document ID: US 5999000 A

L8: Entry 12 of 32 File: USPT Dec 7, 1999

US-PAT-NO: 5999000
DOCUMENT-IDENTIFIER: US 5999000 A

TITLE: Radio-frequency coil and method for resonance imaging/analysis

DATE-ISSUED: December 7, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 13. Document ID: US 5964705 A

L8: Entry 13 of 32 File: USPT Oct 12, 1999

US-PAT-NO: 5964705
DOCUMENT-IDENTIFIER: US 5964705 A

TITLE: MR-compatible medical devices

DATE-ISSUED: October 12, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Truwit; Charles L.	Wayzata	MN		
Liu; Haiying	Minneapolis	MN		

US-CL-CURRENT: 600/423; 324/318, 600/13

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 14. Document ID: US 5928148 A

L8: Entry 14 of 32 File: USPT Jul 27, 1999

US-PAT-NO: 5928148
DOCUMENT-IDENTIFIER: US 5928148 A

TITLE: Method for performing magnetic resonance angiography over a large field of view using table stepping

DATE-ISSUED: July 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wang; Yi	New York	NY		
Lee; Howard M.	Rye	NY		
Khilnani; Neil M.	New York	NY		

US-CL-CURRENT: 600/420; 324/306, 600/415

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 15. Document ID: US 5928145 A

L8: Entry 15 of 32 File: USPT Jul 27, 1999

US-PAT-NO: 5928145
DOCUMENT-IDENTIFIER: US 5928145 A

TITLE: Method of magnetic resonance imaging and spectroscopic analysis and associated apparatus employing a loopless antenna

DATE-ISSUED: July 27, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ocali; Ogan	Baltimore	MD		
Atalar; Ergin	Columbia	MD		

US-CL-CURRENT: 600/410; 324/307, 324/309, 324/318, 600/411, 600/423

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 16. Document ID: US 5910728 A

L8: Entry 16 of 32 File: USPT Jun 8, 1999

US-PAT-NO: 5910728
DOCUMENT-IDENTIFIER: US 5910728 A

TITLE: Simultaneous acquisition of spatial harmonics (SMASH): ultra-fast imaging with radiofrequency coil arrays

DATE-ISSUED: June 8, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel Kevin	Cambridge	MA		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 17. Document ID: US 5905378 A

L8: Entry 17 of 32 File: USPT May 18, 1999

US-PAT-NO: 5905378
DOCUMENT-IDENTIFIER: US 5905378 A

TITLE: Flexible lightweight attached phased-array (FLAP) receive coils

DATE-ISSUED: May 18, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Giaquinto; Randy Otto	Burnt Hills	NY		
Dumoulin; Charles Lucian	Ballston Lake	NY		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 18. Document ID: US 5777474 A

US-PAT-NO: 5777474
DOCUMENT-IDENTIFIER: US 5777474 A

TITLE: Radio-frequency coil and method for resonance imaging/analysis

DATE-ISSUED: July 7, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 19. Document ID: US 5699801 A

L8: Entry 19 of 32 File: USPT Dec 23, 1997

US-PAT-NO: 5699801
DOCUMENT-IDENTIFIER: US 5699801 A

TITLE: Method of internal magnetic resonance imaging and spectroscopic analysis and associated apparatus

DATE-ISSUED: December 23, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Atalar; Ergin	Columbia	MD		
Bottomley; Paul A.	Columbia	MD		
Zerhouni; Elias A.	Baltimore	MD		

US-CL-CURRENT: 600/410; 324/318, 324/322, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 20. Document ID: US 5680047 A

L8: Entry 20 of 32 File: USPT Oct 21, 1997

US-PAT-NO: 5680047
DOCUMENT-IDENTIFIER: US 5680047 A

TITLE: Multipl-tuned radio frequency coil for simultaneous magnetic resonance imaging and spectroscopy

DATE-ISSUED: October 21, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Hts	OH		
Liu; Haiying	Euclid	OH		
Elek; Robert A.	Chardon	OH		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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21. Document ID: US 5572130 A

L8: Entry 21 of 32 File: USPT Nov 5, 1996

US-PAT-NO: 5572130
DOCUMENT-IDENTIFIER: US 5572130 A

TITLE: Method and apparatus for the production of an NMR tomography image using an array of surface coils and multiplexers

DATE-ISSUED: November 5, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Ratzel; Dieter	Rheinstetten			DEX

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

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22. Document ID: US 5565779 A

L8: Entry 22 of 32 File: USPT Oct 15, 1996

US-PAT-NO: 5565779
DOCUMENT-IDENTIFIER: US 5565779 A

TITLE: MRI front end apparatus and method of operation

DATE-ISSUED: October 15, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Arakawa; Mitsuaki	Hillsborough	CA		
Chang; Hsu	Fremont	CA		
Van Heteren; John	San Francisco	CA		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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23. Document ID: US 5560360 A

L8: Entry 23 of 32 File: USPT Oct 1, 1996

US-PAT-NO: 5560360
DOCUMENT-IDENTIFIER: US 5560360 A

TITLE: Image neurography and diffusion anisotropy imaging

DATE-ISSUED: October 1, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Filler; Aaron G.	Seattle	WA		
Tsurda; Jay S.	Mercer Island	WA		
Richards; Todd L.	Seattle	WA		
Howe; Franklyn A.	London			GB2

US-CL-CURRENT: 600/408; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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24. Document ID: US 5548218 A

L8: Entry 24 of 32 File: USPT Aug 20, 1996

US-PAT-NO: 5548218
DOCUMENT-IDENTIFIER: US 5548218 A

TITLE: Flexible RF coils for MRI system

DATE-ISSUED: August 20, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lu; Dongfeng	Williston Park	NY		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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25. Document ID: US 5483158 A

L8: Entry 25 of 32 File: USPT Jan 9, 1996

US-PAT-NO: 5483158
DOCUMENT-IDENTIFIER: US 5483158 A

TITLE: Method and apparatus for tuning MRI RF coils

DATE-ISSUED: January 9, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
van Heteren; John	San Francisco	CA		
Arakawa; Mitsuaki	Hillsborough	CA		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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26. Document ID: US 5461314 A

L8: Entry 26 of 32 File: USPT Oct 24, 1995

US-PAT-NO: 5461314
DOCUMENT-IDENTIFIER: US 5461314 A

TITLE: MRI front end apparatus and method of operation

DATE-ISSUED: October 24, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Arakawa; Mitsuaki	Hillsborough	CA		
Chang; Hsu	Fremont	CA		
Van Heteren; John	San Francisco	CA		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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27. Document ID: US 5086275 A

L8: Entry 27 of 32 File: USPT Feb 4, 1992

US-PAT-NO: 5086275
DOCUMENT-IDENTIFIER: US 5086275 A

TITLE: Time domain filtering for NMR phased array imaging

DATE-ISSUED: February 4, 1992

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Roemer; Peter B.	Schenectady	NY		

US-CL-CURRENT: 324/309; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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28. Document ID: US 4947121 A

L8: Entry 28 of 32 File: USPT Aug 7, 1990

US-PAT-NO: 4947121
DOCUMENT-IDENTIFIER: US 4947121 A

TITLE: Apparatus and method for enhanced multiple coil nuclear magnetic resonance (NMR) imaging

DATE-ISSUED: August 7, 1990

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hayes; Cecil E.	Wauwatosa	WI		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 29. Document ID: US 4885541 A

L8: Entry 29 of 32 File: USPT Dec 5, 1989

US-PAT-NO: 4885541
DOCUMENT-IDENTIFIER: US 4885541 A

TITLE: Apparatus and method for enhanced multiple coil nuclear magnetic resonance (NMR) imaging

DATE-ISSUED: December 5, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hayes; Cecil E.	Wauwatosa	WI		

US-CL-CURRENT: 324/322; 330/286, 330/53, 333/32

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 30. Document ID: US 4881034 A

L8: Entry 30 of 32 File: USPT Nov 14, 1989

US-PAT-NO: 4881034
DOCUMENT-IDENTIFIER: US 4881034 A

TITLE: Switchable MRI RF coil array with individual coils having different and overlapping fields of view

DATE-ISSUED: November 14, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Kaufman; Leon	San Francisco	CA		
Arakawa; Mitsuaki	Hillsborough	CA		
McCarten; Barry M.	Los Altos	CA		
Fehn; John H.	El Sobrante	CA		
Krasnor; Stephen	Oakland	CA		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 31. Document ID: US 4881032 A

L8: Entry 31 of 32 File: USPT Nov 14, 1989

US-PAT-NO: 4881032
DOCUMENT-IDENTIFIER: US 4881032 A

TITLE: Method of, and apparatus for, NMR spectroscopic metabolite imaging and quantification

DATE-ISSUED: November 14, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Bottomley; Paul A.	Clifton Park	NY		
Roemer; Peter B.	Schenectady	NY		
Edelstein; William A.	Schenectady	NY		
Mueller; Otward M.	Ballston Lake	NY		

US-CL-CURRENT: 324/309; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 32. Document ID: US 4825162 A

L8: Entry 32 of 32 File: USPT Apr 25, 1989

US-PAT-NO: 4825162
DOCUMENT-IDENTIFIER: US 4825162 A

TITLE: Nuclear magnetic resonance (NMR) imaging with multiple surface coils

DATE-ISSUED: April 25, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Roemer; Bernard	Schenectady	NY		
Edelstein; William A.	Schenectady	NY		

US-CL-CURRENT: 324/318; 324/312

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Term	Documents
SECOND.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3945801
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THIRDS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	27020
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SEC.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	280401
TERTIARY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	140103
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L9: Entry 1 of 4 File: USPT Nov 21, 2000

US-PAT-NO: 6150816
DOCUMENT-IDENTIFIER: US 6150816 A

TITLE: Radio-frequency coil array for resonance analysis

DATE-ISSUED: November 21, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

ASSIGNEE-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY	TYPE CODE
Advanced Imaging Research, Inc.	Cleveland	OH			02

APPL-NO: 9/ 028336 [PALM]
DATE FILED: February 24, 1998

PARENT-CASE:
This application is based on provisional application No. 60/039,152, filed Feb. 25, 1997, the entire disclosure of which is incorporated herein by reference.

INT-CL: [7] G01 V 3/00

US-CL-ISSUED: 324/318; 324/322
US-CL-CURRENT: 324/318; 324/322

FIELD-OF-SEARCH: 324/318, 324/322, 324/321, 324/300, 324/314, 324/307, 324/309, 600/420, 600/421

PRIOR-ART-DISCLOSED:

U.S. PATENT DOCUMENTS

Search Selected | Search ALL

PAT-NO	ISSUE-DATE	PATENTEE-NAME	US-CL
4398148	August 1983	Barjhoux et al.	324/307
4411270	October 1983	Damadian	128/653
4692705	September 1987	Hayes	324/318
4769605	September 1988	Fox	324/322
4783641	November 1988	Hayes et al.	333/219
4793356	December 1988	Misic et al.	128/653
4799016	January 1989	Rezvani	324/318
4820985	April 1989	Eash	324/318
4825162	April 1989	Roemer et al.	324/318
4833409	May 1989	Eash	324/318
4943775	July 1990	Boskamp et al.	324/322
5057778	October 1991	Rath	324/322
5144240	September 1992	Mehdizadeh et al.	324/318
5194811	March 1993	Murphy-Boesch et al.	324/322
5202635	April 1993	Srinivasan et al.	324/322
5208534	May 1993	Okamoto et al.	324/309
5212450	May 1993	Murphy-Boesch et al.	324/322
5256971	October 1993	Boskamp	324/318
5258717	November 1993	Misic et al.	324/318
5270656	December 1993	Roberts et al.	324/318
5382903	January 1995	Young	324/318
5521506	May 1996	Misic et al.	324/322
5543711	August 1996	Srinivasan et al.	324/318
5548218	August 1996	Lu	324/318
5592083	January 1997	Magnuson et al.	324/300
5592088	January 1997	Matsunaga et al.	324/318
5594338	January 1997	Magnuson	324/318
5602479	February 1997	Srinivasan et al.	324/318
5646531	July 1997	Renz	324/318
5680047	October 1997	Srinivasan et al.	324/318
5777474	July 1998	Srinivasan	324/318
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Michael Burl, Ian R, Young, Examples of the Design of Screened and Shielded RF Receiver Coils, pp. 326-330.

Srinivasan, Improved Radio-Frequency Coil and M Ethod for Resonance/Imaging Analysis, U.S. Patent Application No. 08/993,932, filed Dec. 18, 1997.
International Search Report related to PCT Patent Application No. PCT/US98/03529 dated Jul. 16, 1998.
"Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck" by C. Leussler SMR 1993.
"A Comprehensive Analysis for Estimating Modes in Coupled Resonators"; by Ravi Srinivasan and Haiying Liu.
"Examples of the Design of Screened and Shielded RF Receiver Coils"; Michael Burl and Ian R. Young, pp. 326-330.

ART-UNIT: 282
PRIMARY-EXAMINER: Arana; Louis
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ABSTRACT:

An RF coil array which includes first and second RF coils that are overlapped to eliminate their coupling (to maintain zero mutual inductance between them) through space, a third coil connecting the first and second coils such that there is no net coupling between the first two coils through the third coil, and in which all three coils are well isolated from one another at the resonance frequency or frequencies of interest.

33 Claims, 40 Drawing figures

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Search Results - Record(s) 1 through 4 of 4 returned.

☐ 1. Document ID: US 6150816 A

L9: Entry 1 of 4 File: USPT Nov 21, 2000

US-PAT-NO: 6150816
DOCUMENT-IDENTIFIER: US 6150816 A

TITLE: Radio-frequency coil array for resonance analysis

DATE-ISSUED: November 21, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 2. Document ID: US 5910728 A

L9: Entry 2 of 4 File: USPT Jun 8, 1999

US-PAT-NO: 5910728
DOCUMENT-IDENTIFIER: US 5910728 A

TITLE: Simultaneous acquisition of spatial harmonics (SMASH): ultra-fast imaging with radiofrequency coil arrays

DATE-ISSUED: June 8, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Sodickson; Daniel Kevin	Cambridge	MA		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 3. Document ID: US 5680047 A

L9: Entry 3 of 4 File: USPT Oct 21, 1997

US-PAT-NO: 5680047

DOCUMENT-IDENTIFIER: US 5680047 A

TITLE: Multipl-tuned radio frequency coil for simultaneous magnetic resonance imaging and spectroscopy

DATE-ISSUED: October 21, 1997

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Richmond Hts	OH		
Liu; Haiying	Euclid	OH		
Elek; Robert A.	Chardon	OH		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 4. Document ID: US 5086275 A

L9: Entry 4 of 4 File: USPT Feb 4, 1992

US-PAT-NO: 5086275
DOCUMENT-IDENTIFIER: US 5086275 A

TITLE: Time domain filtering for NMR phased array imaging

DATE-ISSUED: February 4, 1992

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Roemer; Peter B.	Schenectady	NY		

US-CL-CURRENT: 324/309; 324/318, 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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IDENTICALE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
IDENTICALF.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
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DOCUMENT-IDENTIFIER: US 6150816 A
TITLE: Radio-frequency coil array for resonance analysis

Abstract Paragraph Left (1):
An RF coil array which includes first and second RF coils that are overlapped to eliminate their coupling (to maintain zero mutual inductance between them) through space, a third coil connecting the first and second coils such that there is no net coupling between the first two coils through the third coil, and in which all three coils are well isolated from one another at the resonance frequency or frequencies of interest.

Brief Summary Paragraph Right (1):
The present invention relates to resonance systems, such as magnetic resonance imaging (MRI), nuclear quadrupole resonance (NQR), electron spin resonance (ESR) systems, and more particularly to a radio-frequency (RF) coil array and method for use in such systems.

Brief Summary Paragraph Right (2):
In MRI systems or nuclear magnetic resonance (NMR) systems, a static magnetic field B.sub.o is applied to the body under investigation to define an equilibrium axis of magnetic alignment in the region of the body under examination. An RF field is then applied in the region being examined in a direction orthogonal to the static field direction, to excite magnetic resonance in the region, and resulting RF signals are detected and processed. Generally, the resulting RF signals are detected by RF coil arrangements placed close to the body. See for example, U.S. Pat. No. 4,411,270 to Damadian and U.S. Pat. No. 4,793,356 to Misic et al. Typically, such RF coils are either surface type coils or volume type coils, depending on the particular application. Normally separate RF coils are used for excitation and detection, but the same coil or array of coils may be used for both purposes. For multiple surface RF coils for use in NMR, see U.S. Pat. No. 4,825,162 to Roemer, et al.

Brief Summary Paragraph Right (3):
A further increase in S/N can be realized with the use of quadrature coils as compared to the conventional linear coil designs. See for example U.S. Pat. No. 4,467,282 to Siebold and U.S. Pat. No. 4,707,664 to Fehn. Also see U.S. Pat. Nos. 4,783,641 and 4,692,705 to Hayes for a quadrature volume coil, commonly referred to as the "birdcage" coil in the NMR community. For the use of multiple volume coils for use in NMR, see U.S. Pat. No. 5,258,717 to Misic, et al., and the reference article by Leussler for the use of multiple volume coils for simultaneous head and neck imaging (See, C. Leussler, "Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck", SMRM 12th Annual Meeting, New York, Book of Abstracts, page 1349, 1993). Also, reference is made to commonly assigned U.S. patent application Ser. No. 08/745,893 filed on Nov. 8, 1996 titled "Radio-Frequency Coil and Method for Resonance Imaging/Analysis", and Ser. No. 08/993,932 entitled "Improved Radio-Frequency Coil and Method for Resonance Imaging/Analysis", filed on Dec. 18, 1997, the disclosures of which are incorporated herein by reference, for the use of multiple volume and surface coils for use in NMR imaging.

Brief Summary Paragraph Right (4):
The recent introduction of array coils to NMR, has led to commercially available cervical-thoracic-lumbar (CTL) array coil for entire spine imaging, and flexible body arrays for torso imaging. These multichannel coils significantly help reduce scan times. A routine MR study takes approximately 45 minutes, including the patient placement. This is uncomfortable especially for claustrophobic patients in general. In addition, prolonged scan times make the contrast-enhanced studies even more

difficult to obtain. The almost 1 hour MR study with and without the contrast agent makes MR not so suitable for imaging emergency trauma cases.

Brief Summary Paragraph Right (5):
This necessitates a new array coil with high S/N, that will allow the MR study of the torso, head, spine or joints such as the knee, wrist and shoulder etc., to be performed in reduced scan times. This will significantly reduce patient discomfort and increase patient throughput in a MR scanner. The reduced scan times will also allow MRI systems to be used in scanning emergency trauma patients.

Brief Summary Paragraph Right (7):
NQR is a technique that is capable of locating and uniquely identifying nitrogen for the detection of explosives and/or narcotics, even when contained and concealed by other materials. NQR has potential application in general and medical imaging and industrial measurements, in addition to the detection of either explosives (including land mines) or narcotics. See U.S. Pat. Nos. 5,594,338 and 5,592,083 for the design of an RF coil employed in the NQR system.

Brief Summary Paragraph Right (9):
Generally, NQR frequencies of quadrapole nuclei lie within 0.5-5 MHz range. However, for organic chlorine compounds, .sup.35 Cl chemical shifts range from 16-55 MHz. The chemical shift of chlorinated hydrocarbons occurs between 32-45 MHz. This is a very wide frequency range for one single turn RF coil of the aforementioned '338 and '083 references to cover.

Brief Summary Paragraph Right (10):
Even the 0.5 to 5 MHz (a ten fold frequency) range of detection for .sup.14 N in explosives and or narcotics mandate a capacitance of a factor of 100 (f.sup.2 .varies.1/LC) to tune the coil from 5 to 0.5 MHz range, which are overwhelmingly large range of capacitances required to tune the RF coil. Since, the same RF coil was used for a wide frequency range, the RF coil design was un-optimized for the several frequency ranges of operation. This may affect the performance of the RF coil (Q values) and the entire NQR system (transmitter power, S/N), in the detection of low levels of nitrogen compounds found in plastic explosives and narcotics.

Brief Summary Paragraph Right (11):
This necessitates that the RF coil design be optimized for maximum S/N over at least a majority of the frequency ranges of NQR operation and detection in reduced examination times.

Brief Summary Paragraph Right (12):
Even after several years following the introduction of array coils to NMR, the only coil that is commercially used for scanning the human head in a horizontally oriented B.sub.o magnetic field is the quadrature birdcage coil of Hayes '705. Other applications of this coil design are for the whole body, knee and wrist imaging. A birdcage coil consists of two rings connected by several straight segments referred to as legs. A planar schematic of an eight leg high-pass birdcage is shown in FIG. 1a. This coil consists of two end rings R1 and R2 and 8 legs 1 through 8. Each ring section between two legs are interrupted by two series 2C value capacitors. Their combined effect is one capacitor of C value. FIG. 1b is the front view of the birdcage describing the location of the ring with respect to the legs and includes the mode orientation. FIG. 1c is the side view of the coil outline shown for brevity.

Brief Summary Paragraph Right (13):
The birdcage which is of the distributed inductance-capacitance type structure has several frequency modes. Of interest is the first or principal or k=1 quadrature mode. This k=1 quadrature mode has two linear components (1a, 1b), oriented orthogonal to one another as shown in FIG. 1b. As mentioned above, the quadrature coil provided a 41% improvement over the conventional linear coil designs. The birdcage expended half the power when compared to the conventional linear coil, thus significantly reduced the RF power deposited in the patient. The higher order or k>1 modes had no net field at the coil center and generally were not used for imaging. At the k=1 mode, the currents in the coil were cosinusoidally distributed such that the resultant field displayed a homogeneous distribution over the imaging field-of-view (FOV). It is for these regions this coil gained popularity in the NMR community for the several volumetric applications (torso, head, knee, wrist, etc.).

Brief Summary Paragraph Right (14):

The dashed lines of FIGS. 1a, 1b and 1c are planes of symmetry for this birdcage. From FIG. 1b, there are four such planes (I, II, III, IV), that are distributed azimuthally (due to symmetry). There is one additional axial plane (V) that is centrally located between the two end rings R1 and R2, dissecting the coil axis (see FIG. 1c) which, in addition is also a virtual ground plane. The points where the planes of symmetry intersect the birdcage are "a, b, c, d, e, f, g, h" on ring R1 and "i, j, k, I, m, n, o, p" on ring R2, "q, r, s, t, u, v, w, x" on legs 1, 2, 3 . . . 8 respectively of FIG. 1a. Since points "q-x" are located on the virtual ground plane, these points are at virtual ground potential or have no net potential.

Brief Summary Paragraph Right (16):
However, should the virtual ground points "q-x" in the legs of FIG. 1a be shorted, this will result in the coil of FIG. 1e. This will give rise to a new RF gradient mode, bi-phasic in nature with + & - lobes along the coil axis. However, it is noted that a RF gradient mode for the coil of FIG. 1e, has no net field at the coil center (i.e., the RF gradient mode has no net field in the central virtual ground plane of FIG. 1c). Therefore, although FIG. 1e has two birdcages that share one end ring R.sub.12 and even a new mode is realized, no net increase in S/N at the coil center is realized.

Brief Summary Paragraph Right (17):
A quadrature, 3-channel head coil was described by the inventor in previously filed Ser. No. 08/993,932, which provided improved S/N at the coil center and toward the top of the head (see FIG. 2). The coil consisted of two birdcages (coils #1, #2), one distributed, quadrature modified surface coil (coil #3) and passive circuits were used for decoupling individual coils and to minimize the cross-talk between all coils in the array. The coil was operated in the multiple operating modes, with focus to the upper or lower portions of the brain or for routine head studies in one clinical setting, with high S/N and without compromising homogeneity. Here the birdcage, coil #2 and the quadrature surface coil #3 were asymmetrically overlapped and therefore isolated from one another and is the subject of previously filed U.S. Ser. No. 08/745,893.

Brief Summary Paragraph Right (18):
The combination of coils #2 and #3 was then overlapped with birdcage coil #1. Since all three coils in the array were physically separated from one another, and were overlapped to maintain minimal mutual coupling, each coil in the array maintained their own RF current distribution and mode orientation. Several passive coil-to-coil decoupling electronics helped minimize the residual cross-talk between coils in the array. Each quadrature coil signal was routed to individual low noise figure, high gain preamplifiers before digitization. Diode protection circuits were inserted between the coil and the respective preamplifier for preamplifier protection during whole body transmit.

Brief Summary Paragraph Right (19):
A quadrature, 2-channel birdcage array was described in C. Leussler, "Optimized Birdcage Resonators for Simultaneous MRI of Head and Neck", SMRM 12th Annual Meeting, New York, Book of Abstracts, page 1349, 1993 for simultaneous head and neck imaging (see FIG. 3). This coil involved 2 birdcages; a coil 10 for the head and a coil 12 for the neck. The head birdcage 10 had eight fold symmetry, whereas the neck birdcage 12 had only a four-fold symmetry. The neck birdcage 12 had shoulder cut-outs for accomodating the neck as shown in FIG. 3. This coil provided an extended FOV without significantly compromising S/N and homogeneity over the extended FOV. Nevertheless, no increase in S/N was realized over extended FOVs. That is, the S/N of the array coil was comparable to individual head or neck coils over the head & neck scan volume.

Brief Summary Paragraph Right (23):
A multiple surface coil arrangement disclosed in U.S. Pat. No. 5,256,971 to Boskamp is shown in FIG. 4d. Here, two surface coils of similar dimension are overlapped for minimum mutual inductance from one another. A third coil is added to this set, such that the third coil is magnetically isolated from the first and second coils. Here, all three coils are mutually isolated from another. In doing so, the third coil has a different coil geometry than the first and second coils, and extends beyond the FOV of the first and second coils combined.

Brief Summary Paragraph Right (24):
Should this arrangement of coils be flexed around the human torso, the isolation between the third coil and the first and second coils will change, which in turn

will affect the isolation between the first and second coils, as the first and second coils will now start to couple via the third coil. That is, should the third coil begin to couple to either the first or second coils, all coils in the array will begin to couple with each other. This was not satisfactory.

Brief Summary Paragraph Right (28):
Shorting the two virtual ground points of FIG. 5a will result in FIG. 5c. This will give rise to a new RF gradient mode along the coil axis. It will be noted that a RF gradient mode has no net field at the coil center (i.e. the RF gradient mode has no net field in the central virtual ground plane of FIG. 5b). Therefore, although FIG. 4c has two solenoid coils sharing the two virtual ground point "a, b" of FIG. 5a and even a new gradient mode is realized, the homogeneous mode of FIG. 5a will not be affected and no net increase in S/N is realized at the coil center.

Brief Summary Paragraph Right (29):
A single turn solenoid coil of FIG. 6 was used to detect the .sup.14 N signals in crystalline form for detecting concealed explosives and narcotics employing nuclear quadrupole resonance (NQR). See U.S. Pat. Nos. 5,594,338 and 5,592,083 for the design of an RF coil employed in the NQR system.

Brief Summary Paragraph Right (30):
FIG. 6 has one single turn RF coil which is tuned to a wide range (approx 0.5 to 5 MHz), by simply adding large and small value capacitances for coarse and fine tuning with the help of relay switches. As seen, the upper frequency range was ten fold of the lower range which mandated a 100 fold change in capacitance to tuned the coil. Since, the same RF coil was used for a wide frequency range, the RF coil design was un-optimized for the several frequency ranges of operation. This may affect the performance (transmitter power, S/N) of the RF coil and the entire NQR system, in the detection of low levels of nitrogen and chlorine compounds found in plastic explosives and narcotics.

Brief Summary Paragraph Right (31):
This necessitates that the RF coil design be optimized for at least a majority of the frequency ranges of NQR operation and detection which will also help in reducing examination times.

Brief Summary Paragraph Right (32):
This RF coil design will allow for at least one optimized coil in the array that will cover a part of the frequency spectrum, such that all coils in the array combined cover the entire frequency spectrum required for detection. This will help reduce the overall scan frequency range per coil and thus allow rapid tuning of coils in the array. This RF coil design may also be designed to allow for multiple tuning of the coils in the array without crosstalk and capable of simultaneous operation, which will help scan the entire frequency range in reduced scan times.

Brief Summary Paragraph Right (34):
The present invention provides an RF coil with high signal-to-noise ratio (S/N) over the imaging or spectroscopic field-of-view (FOV). The RF coil of the present invention enables one to reduce scan times and therefore patient discomfort without significantly compromising image quality. The RF is capable of operating in different FOVs in the multiple operating modes in one or multiple frequencies. Furthermore, the present invention provides a coil array capable of simultaneous operation in at least one frequency range.

Brief Summary Paragraph Right (35):
A primary objective of the present invention is to provide a novel RF coil design with high S/N, capable of array operation in the single or multiple frequencies. Another objective is to provide an array design, which will provide a high combined S/N than any one coil operated alone. Yet another objective is to provide a RF coil capable of simultaneous multiple frequency operation for resonance imaging/spectroscopic analysis. A further objective is to have coils in the array that are well isolated from one another and maintain their individual current distributions and mode orientations irrespective of the shape of the coil.

Brief Summary Paragraph Right (36):
The design of the inventive coil involves first and second RF coils, that are overlapped to eliminate their magnetic coupling (to maintain zero mutual inductance between them) through space, a third coil physically connecting the first and second coils such that there is no net coupling between the first two coils through the

third coil, and all three coils are well isolated from one another at the resonance frequency or frequencies of interest.

Brief Summary Paragraph Right (38):
In the embodiments of the present invention, the third coil has a FOV nearly identical to that of the combined FOV of the first two coils, and overlaps the combined FOV of the first two coils, such that, the S/N of all three coils combined is greater than any one coil operated alone. In other embodiments, several (first+second+third=integrated) such integrated coils are overlapped for minimal mutual inductance and used in an array configuration. The three individual coils in any one integrated design may be tuned to one or more resonance frequencies, for simultaneous use in imaging or spectroscopic analysis. Depending on the imaging FOV, individual coils in the array can be turned OFF or ON by the programmable transmit/receive (T/R) driver in the resonance system.

Brief Summary Paragraph Right (39):
According to one particular aspect of the invention, a radio-frequency (RF) coil array for resonance imaging/analysis is provided. The coil array includes a first RF coil sensitive to RF signals produced during resonance imaging/analysis; a second RF coil located relative to the first RF coil with substantially zero coupling therebetween at a frequency or frequencies of the RF signals; and a third RF coil located relative to the first RF coil and the second RF coil such that there is substantially zero net current flow between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals.

Brief Summary Paragraph Left (5):
3-Channel Distributed Type Coil Head Array

Brief Summary Paragraph Left (6):
2-Channel Birdcage, Head and Neck Array

Drawing Description Paragraph Right (6):
FIG. 2 is a schematic view of an RF coil having two birdcage coils, and modified spoke type quadrature surface coil;

Drawing Description Paragraph Right (7):
FIG. 3 is a perspective view of a quadrature, 2-channel birdcage array;

Drawing Description Paragraph Right (9):
FIGS. 4b and 4c represent the principal and secondary modes, respectively, for the coil of FIG. 4a;

Drawing Description Paragraph Right (15):
FIG. 7a is a simplified illustration of a coil array in accordance with the present invention in which three coils are shown with respective current distributions;

Drawing Description Paragraph Right (17):
FIG. 7c is a flowchart representing exemplary steps for manufacturing a coil array in accordance with the present invention;

Drawing Description Paragraph Right (22):
FIG. 8a is a planar schematic view of another embodiment of the coil array in accordance with the present invention;

Drawing Description Paragraph Right (23):
FIG. 8b is a schematic front view of the coil array of FIG. 8a;

Drawing Description Paragraph Right (24):
FIG. 8c is a side view of the coil outlines of the array of FIG. 8a;

Drawing Description Paragraph Right (25):
FIG. 8d is a schematic view of a 4-channel, quadrature head array in accordance with the present invention;

Drawing Description Paragraph Right (26):
FIGS. 9a and 9b are front and side views of a 3-channel, quadrature knee array embodiment in accordance with the present invention;

Drawing Description Paragraph Right (27):
FIG. 9c schematically represents a modified knee array in accordance with the present invention;

Drawing Description Paragraph Right (28):
FIGS. 10a and 10b are, respectively, schematic front and side views of a 3-channel, quadrature wrist array in accordance with the present invention;

Drawing Description Paragraph Right (29):
FIG. 11a is a planar schematic view of a 3-channel, quadrature head and neck array in accordance with the present invention;

Drawing Description Paragraph Right (30):
FIG. 11b is a front view of the coil array of FIG. 11a;

Drawing Description Paragraph Right (32):
FIG. 12a is a schematic view of a distributed coil array for spine or torso imaging in accordance with the present invention;

Drawing Description Paragraph Right (33):
FIG. 12b is a schematic view of a modification of the coil array of FIG. 12a in which a fourth coil is formed;

Drawing Description Paragraph Right (34):
FIG. 12c is a schematic view of multiple arrays of FIGS. 12a or 12b integrated together in accordance with the present invention;

Detailed Description Paragraph Right (2):
Referring initially to FIG. 7a, the invention includes first (coil #1) and second (coil #2) RF coils, that are overlapped to isolate the coils from each other, by causing the net shared flux between the coils to be zero. A third RF coil (coil #3) of FIG. 7a, is superimposed on the combination of coils #1 and #2, and physically connects coils #1 and #2 at several points along the coil periphery (see FIG. 7b). For the sake of explanation, the coil #3 may connect to coils #1 & #2 at points A, B and A', B' respectively. This however is done such that there is no net coupling between coils #1 and #2 through coil #3. Thus coils #1 and #2 are isolated from one another and still maintain their individual current distributions and B field orientations. Also, the currents in coil #3 are undisturbed and maintain their original distribution and B field orientation. Thus, all three coils are well isolated from one another and perform the intended function in a resonance experiment independent of the other.

Detailed Description Paragraph Right (3):
Only coil outlines are shown in FIG. 7a for brevity. Not shown are impedances (inductances & capacitances) needed to resonate the RF coil at the frequencies of interest. It will be appreciated that the individual coils of FIGS. 7a may be of the volume type or the surface type or their combination as is discussed more fully below in connection with the specific embodiments.

Detailed Description Paragraph Right (4):
Individual current distributions of coils #1, #2 and #3 are shown in FIG. 7a. Accordingly, their resultant B field orientations are directed in to the plane of the paper (if the fingers of the right hand are curled in the direction of the current, according to the right hand rule, the resultant B field direction of the coil is in the direction of the thumb, and in this case will be pointing in to the plane of the paper). By way of overlapping coils #1 and #2 are isolated from one another and maintain their individual current directions and preferred mode orientations. Here mode is referred to as the frequency mode of interest for a resonance experiment. For the cases of NMR or NQR, the modes of interest may be for one or more distinct radio-frequencies. However, mode orientations are the orientations of the B field over the imaging or the spectroscopic field-of-view (FOV) for individual RF coils, at the frequencies of interest. That is, for a linear RF coil case, there exists one mode that is of interest and one mode orientation. However, for a quadrature RF coil case, there exists two linear modes that are oriented orthogonal to one another. These two linear modes however may be tuned to the same frequency resulting in a quadrature coil, or may be tuned to two distinct frequencies thus depicting a dual tuned RF coil with linear operation at both frequencies.

Detailed Description Paragraph Right (5):
Although it is preferred that coils #1 and #2 maintain identical coil dimensions, it is not an absolute necessity. In fact included in this disclosure is a head and neck array, where coils #1 and #2 are not identical in dimension. Nevertheless for the sake of simplicity, coils #1 and #2 of FIG. 7b have identical dimensions. Coil #3 connected to this combination of coils #1 and #2, encompasses a larger FOV covered by coil #1 or #2 alone. In fact, the FOV of coil #3 in this case is not only comparable but also superimposes the combined FOV of coils #1 and #2.

Detailed Description Paragraph Right (6):
The combined S/N of coils #1 and #2, at the coil center (at the region of overlap) may be close or equal to the S/N of coil #3 of FIG. 7a. Thus the combined S/N of all three coils, that are isolated from another will be substantially greater than any one coil in the array. This is because the direction of currents in all three coils will remain the same and have similar B field orientations. Hence signals from all three coils add up. Since they are isolated from one another the noises from the coils in the array are uncorrelated, resulting in a substantial increase in combined S/N. For details of the mathematical expressions of combined S/N, refer to equations 19 and 20 of U.S. Pat. No. 4,825,162 of Roemer et al.

Detailed Description Paragraph Right (8):
In steps S1 and S2, the first coil #1 is built and tested individually. Next, in step S3 the second coil #2 is built. In step S4, coil #1 and #2 are overlapped to cancel their coupling. Namely, in step S5 it is determined whether coils #1 and #2 are isolated by a predefined acceptable amount (e.g., coupled by less than -20 dB). If no in step S5, the coils #1 and #2 are repositioned relative to each other in an effort to improve the isolation therebetween. Steps S4 and S5 can then be carried out until acceptable isolation is achieved. If yes in step S5, the combined coils #1 and #2 may be tested as represented in step S6.

Detailed Description Paragraph Right (9):
Then coil #3 will be built and added to this assembly as represented in step S7. After this addition, should the isolation between coils #1 & #2 deteriorate as determined in step S8, then either coils #1 & #2 be overlapped to compensate for the cross-talk introduced by the addition of coil #3 or the mechanism of FIG. 7d be used or a combination of both can be used to reisolate coils #1 & #2 after the addition of coil #3. (Step S9). Overlapping coils #1 & #2 again will cancel the net mutual flux shared by coils #1 & #2 after the introduction of coil #3. Final testing of the assembled array can then be carried out in step S10 upon achieving acceptable isolation between the respective coils. Once this optimum overlap is determined, a relatively high precision of duplication can be achieved from one coil batch to another in mass production, by etching the two coils on one or both sides of a single printed circuit board. However in addition to the above, any cross-talk by way of current flow between coils #1 & #2 via coil #3 can be minimized or eliminated in some instances by following the mechanism of FIG. 7d. Furthermore, if there exists any residual cross-talk, this too can be minimized or eliminated by introducing electrical coupling cancellation networks, one example may be similar to that of U.S. Pat. No. 4,769,605 to Fox.

Detailed Description Paragraph Right (12):
For example (see FIG. 7d), let V1 and V2 be the voltages spanned by the two identical capacitors C1, then point A midway between V1 & V2 will be at a potential V.sub.A =V=(V1+V2)/2. In reality, this is an ideal case where the two capacitors are identical in value. However, since 5% tolerance capacitors are generally used in manufacturing, point A may not always be at potential V. Here, point A can be forced to be at potential V by adding or taking away some capacitance value thereby balancing the potential across the capacitors and forcing the points mid-way between them to be at their average values. In the cases where all coils are etched on to a printed circuit board, then the capacitances on the coil can be slightly altered with the addition or subtraction of either small value fixed or trimmer capacitors across C1 and C2 to isolate the coils in the array.

Detailed Description Paragraph Right (14):
It is by these ways (cancelling net mutual shared flux or isolating with any other scheme, proposed additional isolation scheme of FIG. 7d and/or that of Fox, with the above) the isolation between the above coils can be maintained if all coils were fixed or etched on a rigid or on a flexible printed circuit board. Note, an isolation of -20 dB was set as a target. In actuality, this value can be set to any

other number based on the coil design and expected combined S/N. We set a -20 dB value, as this will relate to a 1% loss in combined S/N from the optimum value obtained at coil overlap for coils of identical dimension. For details of coil isolation values and its relation to S/N, please refer to the article by Tropp et al., in the Review of Scientific Instrumentation, Volume 62, Number 11, November 1991.

Detailed Description Paragraph Right (15):
Finally, although it is advantageous to test the individual coils separately as they are built (like that shown in FIG. 7c), once the optimum settings of the coils and isolation values are engineered and specifications met, then simply final testing of the entire RF coil system (consisting of coils #1, #2 and #3) is advised. This will considerably cut the time and costs incurred in the final production line prior to product shipment. However, the proposed flow-chart is a methodical progression of the coil design which is also designed so to enable a relatively easier debugging or trouble shooting of the coil system when needed.

Detailed Description Paragraph Right (16):
A preferred method of coupling to individual coils in the RF coil array and interfacing to the system is shown in FIG. 7e. Let us assume the case when all coils in the array of FIG. 7b are in quadrature. Then there will be a total of six linear modes, operating at the resonance frequency of interest. These modes are a1 & a2 of coil #1, b1 & b2 of coil #2 and c1 & c2 of coil #3, respectively.

Detailed Description Paragraph Right (17):
Generally, the linear modes of a coil are matched to 50 ohms using balanced matched capacitors (not shown) and connected to quadrature hybrids via baluns. Either 50 ohm "criss-cross" discrete network (X.sub.L =X.sub.C =50 ohms; L=124nH, C=50 pF for approx. 64 MHz or 1 .5 T) or shielded, transmission line networks such as coaxial cable traps tuned to frequencies very close to the resonance frequency may be used as baluns to convert the balanced feed to an unbalanced line (see FIG. 7e). This is done to isolate the coil grounds from the system ground and to prevent leakage of the circulating RE currents on the ground shield of the coaxial cable exitting the system.

Detailed Description Paragraph Right (18):
These networks (discrete or transmission line) are shielded as shown by the dotted lines to minimize their interaction with the whole body transmit field and their interaction with the RE coils themselves. Note, it is not necessary to shield the discrete networks but are shown as the preferred embodiment. However, we prefer that the cable traps be shielded so the fields generated by the traps are contained to within the volume encompassed by their shield. This is also done so the cable trap can be made uni or bi directional, depending the nature of the trap's use. The cable trap shown can be made uni directional by shorting its shield to one side of the coaxial cable shield. Likewise it can be made bi-directional by floating the shield, as shown in FIG. 7e.

Detailed Description Paragraph Right (19):
The cable trap consists of two turns 1" in diameter, is wound on a delrin spindle with groves, using a semi-rigid cable of 0.085" o.d. along with fixed and variable capacitors for tuning a specified frequency range centered around the resonance frequency of operation. The RE shield is approximately 1.5.times.1.5.times.0.5" in dimension.With inductor Q values (of that of L in the discrete or that created by the coaxial cable in the transmission line network) of approximately 175-200, impedances (with zero reactances=resistance) of approximately 8-12K.OMEGA. can be realized across the baluns, which is adequate to isolate the grounds at 64 MHz (Resistance R=Q.omega.L). Thus coil to ground leakage and coil to coil interactions be minimized or eliminated and high RE coil efficiences can be maintained.

Detailed Description Paragraph Right (20):
Then the linear signals are combined using a phase shifting network to create a single quadrature output per coil. This is followed by a diode protection network before the preamplifier. All three coils are actively decoupled during whole body transmit (circuitry not shown). Although this decoupling will achieve a-25 dB isolation per coil at the resonance frequency of operation, the additional series-shunt pin-diode protection circuit shown will provide a further -45 dB of isolation between the coil and the preamplifier in every channel. During whole body transmit, diode D1 is turned ON which shunts all the RE present in the signal line to ground before reaching the preamp. Diode D2 is reverse biased during transmit and

helps further isolate the RF present in the signal conductor to the preamp input. During receive D1 is reverse biased and D1 is forward biased to allow all of the RF signals to the preamplifier before digitization and further amplification at the system receiver. Thus the diode circuit will ensure a safe preamp operation (preamp input maximum of roughly +20 dBm).

Detailed Description Paragraph Right (24):
Finally, several of these integrated (coil #1+coil #2+coil #3) coil systems, may be overlapped to provide high combined S/N over extended FOVs, as shown in FIG. 7g.

Detailed Description Paragraph Right (25):
In summary, coils #1, #2 and #3 may be linear or quadrature and of the volume type or surface type or their combination. Should the coils be of the volume type, the dashed line of FIG. 7b will be along the common coil axis. Also, the coils may be tuned to the same or different resonance frequencies. For example, coils #1, #2 and #3 may be tuned to the same resonance frequency. In an another example, coils #1 and #2 may be tuned to one resonance frequency, and coil #3 is tuned to another resonance frequency. In yet another example, the individual coils in the array are tuned to different resonance frequencies and capable of simultaneous operation.

Detailed Description Paragraph Right (30):
FIG. 8c is a side view of the coil outlines, with a head cartoon. As seen, coil #1 coverage extends from the c2-c3 cervical-spine and extends to the top of the cerebellum, coil #2 coverage extends from the mid cerebellum to the top of the head, whereas coil #3 coverage spans the combined FOV's of coils #1 & #2, respectively. Thus, routine head scanning can be accomplished with enhanced S/N, which can be used to reduce scan time or enhance image resolution or a little bit of both can be accomplished with the inventive coil. Furthermore, where specific focus is needed, either coil #1 or coil #2 can be individually turned ON to scan different portions of the human brain. Coupling to the six linear modes of this coil and their interface to the system can be accomplished similar to FIG. 7e. Here, three quadrature coil outputs are interfaced to 3 channels of a NMR system.

Detailed Description Paragraph Right (32):
Coils #1, #2 and #3 have eight legs. Coil #4 is of the self-shielded type and has a total of 16 legs (8 primary and 8 secondary). See the above-mentions application Ser. Nos. 08/745,893 and 08/993,932 for the details of the coil #4's design and construction. Each of the four quadrature outputs from the coils in the head array are interfaced to 4 channels of the resonance receiving system, in this case a NMR receiving system. Please note, all four coils have a shielded, tuned coaxial cable trap in addition to the coupling and interface electronics mentioned in FIG. 7e. These coaxial cable traps help further isolate the RF coil grounds at the preamplifier level to the system ground and interfaces the coil outputs to the system receiver.

Detailed Description Paragraph Right (33):
Please note, individual coils in the array can be turned ON or OFF to image a smaller FOV than the entire coil. If the focus was on the upper parts of the brain, only coils #2 and #4 need be turned ON, whereas if the focus was on the mandible areas only coil #1 may be turned ON.

Detailed Description Paragraph Right (34):
FIG. 9a and 9b are front and side views showing the coil outlines of the knee array. Here, coils #1, #2 and #3 have 4 legs, each. Legs 1, 2, 3 and 4 belong to coils #1 and #2 while 5, 6, 7 and 8 belong to coil #3. All legs are azimuthally distributed as shown in FIG. 9a. Coils #1 and #2 are first overlapped to maintain minimum mutual inductance. Coil #3 is then added which physically connects to coils #1 and #2, such that there is no net coupling between coil #1 and #2 via coil #3. A side view of the coil outlines along with a knee cartoon is shown in FIG. 9b.

Detailed Description Paragraph Right (35):
FIG. 9c is a modified knee array. Here, coils #1 and #2 have 8 legs (1,2,3 . . . 8) each, distributed in the fashion shown. These coils are first overlapped to maintain minimum mutual inductance. Coil #3 that physically connects coils #1 and #2 have only 4 legs (9,10,11,12) which are distributed symmetrically. This arrangement is done to image the foot and the ankle along in addition to imaging the knee and the human calf. Please note, the coils of FIGS. 9 may have a split-top to ease the patient access.

Detailed Description Paragraph Right (36):
FIGS. 10a and 11b are front and side views of a 3 channel, quadrature wrist array. Coils #1 and #2 have 4 legs (1,2,3,4) and are overlapped for minimal mutual inductance. Coil #3 that connects coils #1 and #2 has four legs (5,6,7,8). Thus the entire wrist array has a total of 8 legs as shown in FIG. 10a. Please note, the opening of the wrist coil is elliptical in shape to accommodate imaging of the fingers of the human hand. This also facilitates lateral placement of the coil along side the patients body inside a MRI machine. This high S/N coil allows for high-resolution imaging of the carpal ligaments of the human wrist.

Detailed Description Paragraph Right (37):
A planar schematic of the coil is shown in FIG. 11a. Coil #1 has 8 legs (1,2,3 . . . 8) and covers the head FOV. Coil #2 has 8 legs (1,2,3 . . . 8) and has shoulder cut outs to accomodate the entire human neck. Each of these coils are resonant at the NMR frequency. Coil #1 and #2 are overlapped for minimal mutual inductance. Coil #3 connects coil #1 and #2 at eight points and hence has 4 legs distributed at right angles from one another. Thus the entire head and neck coil has 12 legs. Here, coil #1 is resonant with C1, and coil #2 is resonant with C2 whereas coil #3 is resonant with C1, C2 and C3, respectively. A front view of the coil is shown in FIG. 11b.

Detailed Description Paragraph Right (38):
Here, just coil #1 or coil #2 can be turned ON for performing head or neck only studies. Also, all coils (#1, #2 & #3) can be turned ON to perform extended FOV head and neck studies, simultaneously. In this case where coil #3 spans a similar FOV as the combined FOVs of coils #1 & #2, the signals add up and since the noises are uncorrelated, enhanced S/N will be realized, unlike the prior art of FIG. 3. Also, coil #4 of FIG. 8d, can be added to the 3 channel, quadrature coil of FIG. 11 to further improve the S/N toward the top of the head (see FIG. 11c).

Detailed Description Paragraph Right (39):
FIG. 12a is a distributed surface coil array for brain of torso imaging. Here, coil #1 and #2 are overlapped to maintain minimum mutual inductance. Therefore, the net flux shared by these two coils is zero. Both these coils are identical in dimension. They comprise of 2 ring segments, 4 legs and are resonated with C1 value capacitors. These coils are matched to 50 ohms, across the terminals "a, b" similar to the circuit of FIG. 7e and interfaced to 2 channels of the MRI system. Evidently, these two outputs can be matched and summed using a phase shifter, resulting in a single channel quadrature output.

Detailed Description Paragraph Right (40):
Coil #3 consists of 2 ring segments and 3 legs that connect to coils #1 and #2. Coil #3 is tuned to the resonance frequency of interest with C1 and C2 value capacitors. Coil #3 is matched to 50 ohms across "c" terminal using the similar circuitry of FIG. 7e. After the addition of coil #3, the isolation between coils #1 and #2 remained virtually the same, is indicative of a well isolated system. Please note, one integrated RF coil unit I comprises of coils #1, #2 and #3, respectively.

Detailed Description Paragraph Right (41):
FIG. 12b is an extension of FIG. 12a, where coil #3 has an additional ring segment resulting in coil #4 on the same coil system. This coil is tuned to the resonance frequency of interest with C1, C3 and C4. Coils #1 and #2 here are tuned with C1, whereas coil #3 is tuned with C1 and C3. The outputs "a, b" can either be routed to two receiver channels, or combined using a phase shifting network resulting in a single quadrature output. Similarly, the other two outputs, "c, d" can be either routed to two other receiver channels or combined prior to the receiver. Please note, one integrated RF coil unit I comprises of coils #1, #2, #3 and #4, respectively.

Detailed Description Paragraph Right (42):
FIG. 12c is a result of many such integrated RF coil circuits I, II, . . . of FIGS. 12a or 12b, in an array configuration. The coils of FIGS. 12 may be used to image the spine or wrapped around the human torso for imaging the liver, kidney, heart, etc. They may also be used to scan both feet for imaging the blood flow.

Detailed Description Paragraph Right (44):
FIG. 13a consists of a total of 3 solenoid coils. Coils #1 and #2 are identical in dimension. They both have 2 turns separated by a set distance and are tuned with C1 value capacitors. These coils are overlapped to maintain minimum mutual inductance, thus the net flux shared by these two coils are zero. Coil #3 is then introduced by

shorting virtual ground points "a, b" in coil #1 to "c, d" in coil #2. However, the shorting between points "a, c" is done with two turns, the first turn exists in the virtual ground plane of coil #1 and the second turn in the virtual ground plane of coil #2. This is done such that coils #1 and #2 will not see coil #3. Also, this shorting is interrupted with C2 and the shorting between the points "b, d" is interrupted with C3 value capacitor. Please note, only two turns are used for coils #1 and #2, for simplicity. In practice, coils #1 and #2 may have N (N.gtoreq.1) turns, and coil#3 may have M (M=N or M.noteq.N) turns.

Detailed Description Paragraph Right (46):
A preferred embodiment is where all coils are tuned to different NQR frequencies and have their own tunable range of frequencies, lets say for example coil #1 covers from 0.5-1.5 MHz, coil #2 from 1.5-3.0 MHz and coil #3 from 3-5 MHz, respectively. Each coil design is optimized to cover this frequency range and has its own capacitor bank as shown in FIG. 13b to tune the specified frequency range. The individual switches may be computer controlled (not shown) to tune the individual RF coil to the specified resonance frequencies. The object that need to be scanned is introduced along the coil axis.

Detailed Description Paragraph Right (47):
Alternately, the solenoid design may be adapted to a surface type design and may be used for surface detection of drugs, narcotics, explosives, etc. The RF coil may also be used in a quasi surface--volume type design as well, for several medical and non-medical applications.

Detailed Description Paragraph Right (48):
FIG. 14 is a system block diagram, which illustrates the utility of the RF coil of the pressent invention in NMR imaging and spectroscopy, for example. The system has a main magnet which covers the time varying gradient coils, an RF shield that isolates the RF coil from the gradient coils and a whole-body RF coil most commonly used for uniform B field transmit over a large imaging FOV. The main magnet strength sets the NMR frequency of operation. The time varying gradient fields help spatially encode the NMR signals. The RF whole body coil is used to transmit, while the local RF coil is used to pick up the NMR signals from the object under investigation (NMR phantom). A number of receiver coils may be used in an array configuration and may be summed either analog or digitally to produce the resultant image. Signals from the several receiver ports may be acquired via one or multiple receiver channels. An n-to-1 channel multiplexer is shown in the drawing. This helps by-pass n channel coil data to use one channel of the NMR system. Alternatively, an n channel NMR system may also be used.

Detailed Description Paragraph Right (50):
From all the above description, for someone skilled in the art, is must now be apparent that the inventive novel concept of FIG. 7 may be adapted to a number of different coil designs for the several resonance techniques, such as NMR, NQR, etc. It must also now be apparent that the individual coils in an integrated RF coil system may be tuned to the same or different frequencies.

Detailed Description Paragraph Right (51):
It is to be noted that the individual coils in the array may be shaped in such a way to provide a high S/N and uniform coverage over the imaging FOV. The coils may be used to image in the different operating modes. The signal may be combined prior to the preamplifier or post the preamplifier in analog or digital fashion. The individual coils in the array may be tuned to one or more frequencies.

Detailed Description Paragraph Right (52):
It must be further apparent that the coil designs in the distributed cases may be of the low-pass, high-pass, band-pass, band-stop or a combination of the above different configurations. Also, the coils may be of the volume type, surface type or a combination of both. Individual coils in the array may be linear or in quadrature. The coils may be used for transmit only, receive only or may be used for transmit and receive purposes. Individual coils in the array may be interfaced to separate channels in the multi-channel resonance system or may be time-multiplexed to one or more channels of a single or multi-channel resonance system.

Detailed Description Paragraph Left (4):
RF Coil System Arrays

Detailed Description Paragraph Left (5):

Embodiment #2--3 Channel, Quadrature Birdcage Array

Detailed Description Paragraph Left (6):
Embodiment #3--4 Channel, Quadrature Head Array

Detailed Description Paragraph Left (7):
Embodiment #4--3 Channel, Quadrature Knee Array

Detailed Description Paragraph Left (8):
Embodiment #5--3 Channel, Quadrature Wrist Array

Detailed Description Paragraph Left (10):
Embodiment #7--Distributed Surface Coil Array for Spine and Torso

Other Reference Publication (2):
Michael Burl, Ian R, Young, Examples of the Design of Screened and Shielded RF Receiver Coils, pp. 326-330.

Other Reference Publication (3):
Srinivasan, Improved Radio-Frequency Coil and M Ethod for Resonance/Imaging Analysis, U.S. Patent Application No. 08/993,932, filed Dec. 18, 1997.

Other Reference Publication (7):
"Examples of the Design of Screened and Shielded RF Receiver Coils"; Michael Burl and Ian R. Young, pp. 326-330.

CLAIMS:

1. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil electrically connected and located relative to the first RF coil and the second RF coil such that there is substantially zero net current flow between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals.

2. The coil array of claim 1, wherein the first RF coil, second RF coil and third RF coil are sufficiently isolated from one another to maintain predefined current distributions and mode orientations for the respective coils.

3. The coil array of claim 1, wherein the third RF coil has a field-of-view which is similar as compared to a combined field-of-view of the first RF coil and the second RF coil.

4. The coil array of claim 1, wherein each of the first RF coil, second RF coil and third RF coil are volume type coils.

5. The coil array of claim 4, wherein each of the first RF coil, second RF coil and third RF coil are birdcage type coils.

6. The coil array of claim 5, wherein the coil array is sized to receive at least one of a human head, a human knee, and a human wrist within the first RF coil second RF coil and third RF coil.

7. The coil array of claim 5, further comprising a fourth RF coil positioned toward an end of the coil array.

8. The coil array of claim 5, wherein the coil array is sized to receive a human head.

9. The coil array of claim 5, wherein the coil array is sized to receive a human head and neck.

10. The coil array of claim 4, wherein each of the first RF coil, second RF coil and

third RF coil are solenoid type coils.

11. The coil array of claim 10, wherein the coil array is sized to receive at least one of a human head, knee, wrist or torso within the first RF coil, second RF coil and third RF coil.

12. The coil array of claim 1, wherein each of the first RF coil, second RF coil and third RF coil are surface type coils.

13. The coil array of claim 12, further comprising a fourth RF coil.

14. The coil array of claim 1, wherein the net shared magnetic flux between the first RF coil and the second RF coil is substantially zero.

15. The coil array of claim 1, wherein the third RF coil is physically connected to the first RF coil and the second RF coil.

16. The coil array of claim 1, wherein the first RF coil and the second RF coil maintain substantially similar physical dimensions.

17. The coil array of claim 1, wherein each of the first RF coil, second RF coil and third RF coil is configured to provide a quadrature output.

18. The coil array of claim 1, further comprising means for selectively turning the first RF coil, second RF coil and third RF coil on and off to control a mode of operation.

19. The coil array of claim 1, wherein at least one of the first RF coil, second RF coil and third RF coil is a volume type coil, and at least another of the first RF coil, second RF coil and third RF coil is a surface type coil.

20. The coil array of claim 1, wherein the first RF coil, second RF coil and third RF coil are tuned to the same resonance frequency.

21. The coil array of claim 1, wherein the first RF coil and the second RF coil are tuned to one resonance frequency, and the third RF coil is tuned to another resonance frequency.

22. The coil array of claim 1, wherein the first RF coil, second RF coil and third RF coil are tuned to respective different resonance frequencies.

23. A system comprising the coil array of claim 1, and further comprising means for driving the coil array during imaging/analysis.

24. A resonance imaging/analysis system, comprising:

an RF coil as recited in claim 1; and

means for processing RF signals which are at least one of received from the RF coil and transmitted from the RF coil in order to obtain a resonance image/analysis.

25. The coil array of claim 1, wherein the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

26. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil electrically connected and located relative to the first RF coil and the second RF coil such that there is substantially no net coupling between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals.

27. The coil array of claim 26, wherein the isolation between the first and second

RF coils is substantially the same with the third RF coil as without the third RF coil.

28. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil located relative to the first RF coil and the second RF coil such that each of the first RF coil, second RF coil and third RF coil are substantially isolated from the other coils at the frequency or frequencies of the RF signals;

wherein a field-of-view of the third RF coil substantially overlaps and is substantially similar to a combined field-of-view of the first and second RF coils.

29. The coil array of claim 28, wherein the third RF coil is electrically connected to the first RF coil and the second RF coil.

30. The coil array of claim 28, wherein

the third RF coil is electrically connected to the first and second RF coils, and

the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

31. The coil array of claim 30, wherein the third RF coil is electrically connected to the first RF coil and the second RF coil.

32. A radio-frequency (RF) coil array for resonance imaging/analysis, comprising:

a first RF coil sensitive to RF signals produced during resonance imaging/analysis;

a second RF coil located relative to the first RF coil with substantially no net coupling therebetween at a frequency or frequencies of the RF signals; and

a third RF coil located relative to the first RF coil and the second RF coil such that there is substantially no net coupling between the first RF coil and the second RF coil via the third RF coil, each of the first RF coil, second RF coil and third RF coil being substantially isolated from the other coils at the frequency or frequencies of the RF signals;

wherein a field-of-view of the third RF coil substantially overlaps and is substantially similar or larger than a combined field-of-view of the first and second RF coils.

33. The coil array of claim 32, wherein

the third RF coil is electrically connected to the first and second RF coils, and

the isolation between the first and second RF coils is substantially the same with the third RF coil as without the third RF coil.

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L9: Entry 2 of 4 File: USPT Jun 8, 1999

DOCUMENT-IDENTIFIER: US 5910728 A
TITLE: Simultaneous acquisition of spatial harmonics (SMASH): ultra-fast imaging with radiofrequency coil arrays

Abstract Paragraph Left (1):
A magnetic resonance (MR) imaging apparatus and technique exploits spatial information inherent in a surface coil array to increase MR image acquisition speed, resolution and/or field of view. Partial signals are acquired simultaneously in the component coils of the array and formed into two or more signals corresponding to orthogonal spatial representations. In a Fourier embodiment, lines of the k-space matrix required for image production are formed using a set of separate, preferably linear combinations of the component coil signals to substitute for spatial modulations normally produced by phase encoding gradients. The signal combining may proceed in a parallel or flow-through fashion, or as post-processing, which in either case reduces the need for time-consuming gradient switching and expensive fast magnet arrangements. In the post-processing approach, stored signals are combined after the fact to yield the full data matrix. In the flow-through approach, a plug-in unit consisting of a coil array with an on board processor outputs two or more sets of combined spatial signals for each spin conditioning cycle, each directly corresponding to a distinct line in k-space. This partially parallel imaging strategy, dubbed SiMultaneous Acquisition of Spatial Harmonics (SMASH), is readily integrated with many existing fast imaging sequences, yielding multiplicative time savings without a significant sacrifice in spatial resolution or signal-to-noise ratio. An experimental system achieved two-fold improvement in image acquisition time with a prototype three-coil array, and larger factors are achievable with ther coil arrangements.

Brief Summary Paragraph Right (3):
In general, MRI devices establish a constant homogeneous magnetic field to orient nuclear spins, apply a specific additional bias field gradient in a known plane or region under consideration, and apply a radiofrequency pulse or a sequence of pulses to perturb the nuclei. These nuclei in the known bias field gradient emit an RF signal in a specific band determined by the magnetic field distribution, and these RF emissions are detected by receiving coils and stored as a line of information in a data matrix known as the k-space matrix. The full matrix is built up by successive cycles of conditioning the spins, perturbing them, and collecting RF emissions. An image is then generated from this matrix by Fourier transformation, which converts the frequency information present in the RF oscillations to spatial information representing the distribution of nuclear spins in tissue or other imaged material.

Brief Summary Paragraph Right (9):
Several fast imaging schemes have been proposed to date using simultaneous data acquisition in multiple RF coils, as described in: D. Kwiat, S. Einav, G. Navon, A decoupled coil detector array for fast image acquisition in magnetic resonance imaging. Med Phys, 18:251-265 (1991); D. Kwiat, S. Einav, Preliminary experimental evaluation of an inverse source imaging procedure using a decoupled coil detector array in magnetic resonance imaging. Med Eng Phys, 27, 257-263 (1995); J. W. Carlson, T. Minemura, Imaging time reduction through multiple receiver coil data acquisition and image reconstruction. Magn Reson Med 29, 681-688 (1993) and U.S. Pat. No. 4,857,846 of J. W. Carlson; and J. B. Ra, C. Y. Rim, Fast imaging using subencoding data sets from multiple detectors. Magn Reson Med 30, 142-145 (1993). These approaches have offered the promise of significant savings in image acquisition times.

Brief Summary Paragraph Right (11):

The approach of Ra and Rim involves a simultaneous acquisition technique in which images of reduced FOV are acquired in multiple coils of an array and the Nyquist aliasing in those images is undone by reference to component coil sensitivity information. The unaliasing procedure involves a pixel-by-pixel matrix inversion to regenerate the full FOV from multiple copies of the aliased image data. The "subencoding" technique of Ra and Rim relies on estimates of component coil sensitivities by effectively probing the sensitivity at each pixel. This pixel-by-pixel approach can lead to local artifacts; for example, the matrix inversion can begin to fail in regions of low sensitivity. Further, by its very nature as a pixel by pixel dealiasing approach, the Ra & Rim method is computation-intensive and is limited to postprocessing, as all image data must be present before the reconstruction can be undertaken.

Brief Summary Paragraph Right (12):
In a related area, multiple coil signal collection has been used in MR phased array systems as reported in R. B. Roemer, W. A. Edelstein, C. E. Hayes, S. P. Souza, and O. M. Mueller, The NMR phased array. Magn. Reson. Med. 26, 192-225 (1990); C. E. Hayes and P. B. Roemer, Noise correlations in data simultaneously acquired from multiple surface coil arrays. Magn. Reson. Med. 16, 181-191 (1990); C. E. Hayes N. Hattes, and P. B. Roemer, Volume imaging with MR phased arrays. Magn. Reson. Med. 18, 309-319 (1991). The increased information content of the multiple received signals in such systems has been used to increase the signal-to-noise ratio (SNR) of MR images. Since their initial description, phased arrays have seen increasing use in clinical MR imaging. For example, the improvements in SNR provided by phased arrays have allowed significant advances in imaging of the pulmonary vasculature as reported by T. K. F. Foo, J. R. MacFall, C. E. Hayes, H. D. Sostman, and B. E. Slayman, Pulmonary vasculature: single breath-hold MR imaging with phased array coils. Radiology 183, 473-477 (1992). Still, with a few notable exceptions, the bulk of phased array applications have addressed increased sensitivity, with little effort towards improving image acquisition speed or spatial resolution.

Brief Summary Paragraph Right (13):
An MRI system according to the present invention uses a multiple-coil data collection system to acquire some portion of the k-space matrix in parallel, rather than sequentially in time. In a preferred embodiment, signals are obtained from multiple RF coils each occupying a different position with respect to the imaged volume, and each therefore having different but at least partially overlapping spatial sensitivities. The multiple coils are preferably positioned and/or their outputs sampled in a manner to minimize inductive coupling, but they need not individually span the full region of interest nor be fully independent. The signals collected in this plurality of coils are then combined with suitably chosen weights to produce two or more composite signals, each of which approximates a sinusoidal sensitivity modulation, or "spatial harmonic." As used herein, the term "spatial harmonic" refers to a sinusoidal and/or cosinusoidal variation in spatial sensitivity with a wavelength that is an integer fraction of the extent of the field of view. Each line of these spatial harmonic composite signals constitutes an additional line of the k-space matrix which would require a distinct gradient step in a conventional MR acquisition. Thus, the use of such signal combinations eliminates the need for some of the conventional gradient steps. The technique just outlined is referred to herein as SiMultaneous Acquisition of Spatial Harmonics (SMASH), and it may be used to reduce image acquisition times by a multiplicative factor without a significant sacrifice in spatial resolution or signal-to-noise ratio (SNR). The invention also includes non-Fourier embodiments, in which the coil signals are tranformed or combined with weights to yield composite signals which each correspond to a non-Fourier spatial distribution, such as a wavelet.

Brief Summary Paragraph Right (16):
A representative apparatus implementing the invention simultaneously acquires partial signals from multiple coils in a surface coil array and combines them into two or more differently weighted combinations that accurately represent several spatial harmonics. The weights may be generated by theoretical calculations based on coil geometry, or else they may be derived by a calibration protocol carried out on a phantom image or on a pre-scan at the time of in vivo imaging. The calibration protocol typically uses a numerical optimization algorithm to determine coefficients for linear signal combination which will best approximate the desired spatial harmonics. Once the spatial harmonic signal combinations have been formed, the missing lines in k-space are filled in, and the conventional MR image is generated by Fourier transformation of the expanded data matrix. Preferably, the coils in a SMASH coil array are surface coils, i.e. coils positioned on the surface of the body

and designed to capture the MR signal efficiently over a restricted region of interest. Such surface coils typically have a spatial extent on the order of 5-50 centimeters.

Drawing Description Paragraph Right (5):
FIG. 2B illustrates a receiver coil array of the invention and a representative image plane geometry;

Drawing Description Paragraph Right (6):
FIGS. 3, 3A, 3B, 3C, 3D and 3E illustrate another receiver coil array of the invention together with individual sensitivities and the construction of spatial harmonics from component coil sensitivities;

Drawing Description Paragraph Right (7):
FIGS. 4-1, 4-2, and 4-3 illustrate sensitivity reference images formed from the coil array of FIG. 2B and the complex sensitivities of the coils;

Detailed Description Paragraph Right (1):
FIG. 1 illustrates schematically an MRI system 10 which includes the usual static magnet assembly, gradient coils, and transmit RF coils collectively denoted 12 under control of a processor 14, which typically communicates with an operator via a conventional keyboard/control workstation 16. These devices generally employ a system of multiple processors for carrying out specialized timing and other functions in the MRI system 10 as will be appreciated. Accordingly, as depicted in FIG. 1, an MRI image processor 18 receives digitized data representing RF NMR responses from an object region under examination (e.g., a human body 1) and, typically via multiple Fourier transformation processes well-known in the art, calculates a digitized visual image (e.g., a two-dimensional array of picture elements or pixels, each of which may have different gradations of gray values or color values, or the like) which is then conventionally displayed, or printed out, on a display 18a.

Detailed Description Paragraph Right (2):
In such overall operation, the apparatus of the present invention is largely conventional. However, in accordance with the present invention, the basic RF data acquisition is modified, and subsequent signal processing altered, by providing a plurality of surface coils 20a, 20b . . . 20i for simultaneous signal reception, along with corresponding signal processing and digitizing channels. The processor recombines the cllected values into two or more spatial harmonics from which multiple lines of the signal matrix are developed. This recombination may be performed in real time as the data arrives, or after the fact via postprocessing as is convenient with the apparatus and the calibration information at hand.

Detailed Description Paragraph Right (6):
Generally prior art magnetic resonance receiver coils, especially surface coils, do not have uniform sensitivity. Signals from different regions of the imaged volume produce different currents in an RF coil, with the spatial variation in sensitivity being simply related to the inhomogeneity of RF field produced by the coil over the sample volume. For a standard circular surface coil, there is a sensitivity "sweet spot" centered at roughly one diameter below the coil, with a monotonic falloff of sensitivity together with increasing phase differences in all directions. Traditional imaging protocols often position the receiving coil with the target tissue at its region of maximum sensitivity.

Detailed Description Paragraph Right (9):
Thus, if one were to collect a signal with a coil shaped so that the response it picked up had the form of such a spatial harmonic, it would directly yield a shifted k-space entry, and an array of such coils might simultaneously acquire several lines of data corresponding to several distinct gradient steps.

Detailed Description Paragraph Right (10):
Applicant further realized that the sensitivity of each individual coil in an array need not be strictly sinusoidal, so long as the net sensitivity of the array may be configured to assume the desired sinusoidal shape. This realization significantly relaxes the constraints on coil design and disposition, and in conjunction with a calibration process, allows the signals from multiple coils with a wide range of shapes and geometries to be combined in several ways to yield multiple net signals with each with its own distinct sinusoidal sensitivity profile.

Detailed Description Paragraph Right (11):
FIG. 2B illustrates the geometry of an exemplary coil array and a representative image plane above it. A set of three adjacent coils are used with separate coil outputs in a surface array extending generally across the region and planes of interest. The patient is positioned and the spin preparation fields are applied to condition a plane P which, by way of example may intersect the patient's heart to image the blood vessels thereof, or which may intersect a region of the abdomen to image its contents. The coil array 20 is located above or below the region to be imaged, so that each coil 20a, 20b, 20c has at least some sensitivity to RF signals emanating from region P. In the illustrated embodiment the coils are each somewhat overlapped in the y-direction to minimize inductive coupling. As shown, they are overlapped at neighboring edges. Each coil, viewed individually has a sensitivity function which is highest directly above or below the center of the coil, and which falls off with distance from the coil center. In accordance with the present invention, the signals from these coils are combined to produce several separate "virtual" or synthetic signals which each correspond to a pure spatial harmonic.

Detailed Description Paragraph Right (12):
FIG. 3A demonstrates this situation schematically for a set of eight rectangular coils 20a, 20b, . . . 20h laid end-to-end, with a slight overlap. As shown in line (A) of the Figure, each coil 20a, 20b . . . has a sensitivity curve a, b, . . . which rises to a broad peak directly under the coil and drops off substantially beyond the coil perimeter. The sum of the coil sensitivities forms a relatively constant sensitivity, over the full length of the array, corresponding to the zero.sup.th spatial harmonic. The remaining lines (B)-(E) of FIGS. 3B-3E illustrate recombinations of different ones of these individual offset but otherwise identical coil sensitivity functions into a new synthetic sinusoidal spatial sensitivity. Different weightings of the individual component coil sensitivities lead to net sensitivity profiles approximating several of the spatial harmonics of K.sub.y. In the FIGURE, coil sensitivities (modeled schematically for this Figure as Gaussian in shape) are combined to produce harmonics at various fractions of the fundamental spatial wavelength =.lambda..sub.y 2.pi./K.sub.y, with .lambda..sub.y being on the order of the total coil array extent in y. Weighted individual coil sensitivity profiles are depicted as thin solid lines beneath each component coil. Dashed lines represent the sinusoidal or cosinusoidal weighting functions. Combined sensitivity profiles are indicated by thick solid lines. These combined profiles closely approximate ideal spatial harmonics across the array.

Detailed Description Paragraph Right (13):
For a weighted sum of component coil signals, the net sensitivity profile C.sup.tot (x,y) is a linear combination of the intrinsic sensitivity profiles of the component coils. In this case, for an N-component coil array, we have ##EQU1## where n.sub.j is the weighting coefficient of the j.sup.th coil sensitivity function C.sub.j (x,y).

Detailed Description Paragraph Right (14):
In accordance with the aspect of the invention illustrated in FIG. 3, by simply weighting the signals in the coil array with appropriate weights and combining them, one obtains a signal whose amplitude is modulated by the spatial harmonic corresponding to that combination of the coil sensitivities. That is, the N (in this case, eight) component coil signals are acquired simultaneously and then multiplied by various weights and recombined in a total of up to N independent combinations each representing a spatial harmonic corresponding to a distinct gradient phase encoding step or distinct offset in k-space. This additional processing does not add in any way to image acquisition time, and it may be performed quickly on stored signal data as part of a post-processing algorithm or else may be implemented in a simultaneous flow-through fashion as described below.

Detailed Description Paragraph Right (15):
In actual practice, the sensitivity profiles of RF surface coils are not simple Gaussian profiles but more complicated functions which are, in general, complex in the mathematical sense of having both real and imaginary components. The coil sensitivity functions must describe both the magnitudes and the phases of the signals produced by precessing spins at various distances from the coil center, and these magnitudes and phases vary according to the reciprocity relation ##EQU2## where E(r) is the voltage induced in a coil by a given voxel at position ir, m(r) is the nuclear magnetic moment of the voxel, and B.sub.1xy (r) is the xy vector component of the field generated at r by a unit current in the coil as described for example in D. I. Hoult and R. E. Richards, The signal-to-noise ratio of the nuclear

magnetic resonance experiment. J. Magn. Reson. 24, 71-85 (1976).

Detailed Description Paragraph Right (16):
The fact that the sensitivity functions are complex rather than purely real has two main consequences for applicant's SMASH reconstruction: first, applicant assigns complex weights n.sub.j to capture the full information content of the coil sensitivity functions, and second, the target complex exponential sensitivities

Detailed Description Paragraph Right (17):
In setting up an actual MR imaging apparatus, the coil geometry for SMASH signal acquisition may be that of an MR phased array, as described in P. B. Roemer, W. A. Edelstein, C. E. Hayes, S. P. Souza, and O. M. Mueller, The NMR phased array, Magn. Reson. Med. 16, 192-225 (1990); C. E. Hayes and P. B. Roemer, Noise correlations in data simultaneously acquired from multiple surface coil arrays. Magn. Reson. Med. 26, 181-191 (1990); and C. E. Hayes, N. Hattes, and P. B. Roemer, Volume imaging with MR phased arrays. Magn. Reson. Med. 18, 309-319 (1991). Indeed, many of the hardware components useful for SMASH imaging are already present in traditional phased arrays, which contain multiple inductively decoupled coils with some spatial separation, and which include separate receivers for independent collection of data from the coils. The technical problem of minimizing inductive coupling of such coils has been previously addressed for multiple-coil constructions in the context of phased-array MR imaging devices, and two basic strategies were developed. One strategy is to design a coil array with an appropriately chosen overlap of neighboring component coils to minimize inductive coupling. The second strategy involves the use of low input impedance preamplifiers on each component coil channel. Both of these features are advantageously applied in various embodiments of the present invention.

Detailed Description Paragraph Right (18):
Applicant has implemented a basic embodiment of the invention as outlined above in the following manner. First, a linear RF coil array as shown in FIG. 2B was selected having a geometry suitable for spatial harmonic generation. This array was a cardiac imaging array having three 200 mm.times.150 mm rectangular coils adjacent to each other and partially overlapping. The appropriateness of the coil geometry was first tested in numerical simulations, using analytic integration of the Biot-Savart law to calculate the transverse field B.sub.1xy (r) of Eq. [6], and hence to model the sensitivity profile of each coil. Taking into account the number of spatial harmonics that could comfortably be generated using the coil array, and therefore the fraction of the total k-space matrix to be collected, partial image acquisitions were planned, and a regimen of gradient steps and RF pulses were established. Image data were then acquired from each coil simultaneously in the separate channels of the array, using a conventional fast imaging sequence. Since only a fraction of the usual signal data was to be used to generate the final images, only a faction of the usual imaging time was spent on data collection.

Detailed Description Paragraph Right (20):
Determination of the weightings needed for spatial harmonic generation requires knowledge of component coil sensitivities. For the prototype, individual complex sensitivity profiles were determined for each of the component coils in the array using phantom image data instead of numerical simulations. This process is illustrated in FIGS. 4-1 to 4-3 and described in more detail below under the heading of "Determination of coil sensitivity and optimal component coil weights."

Detailed Description Paragraph Right (22):
For simple coil configurations, the sensitivity may be modeled analytically with accuracy, rather than be empirically derived from normalizing the coil responses to a uniform sample. More generally, the process illustrated in FIGS. 4-1 to 4-3 and 4A-1 to 4A-4 described below may be used to determine suitable coefficients for a given image plane, and this process may be repeated to compile a table for each of many planes so that the apparatus need only look up the necessary weights in order to produce the desired composite signals for the plane of interest. When the sensitivity information is known in advance, then, reconstruction does not require a preliminary sensitivity measurement or an iterative fitting procedure, and simply involves a set of weighted sums. In either case, once the new composite signals are formed, the additional data must simply be entered at the correct matrix position, and the full matrix subjected to a fast Fourier transform. This processing is as straightforward as the postprocessing which is commonly performed when conventional images are generated from phased arrays. Furthermore, once the full k-space matrix is constructed using the SMASH technique, subsequent processing of the k-space

matrix to yield the image is identical to that presently employed in substantially all Fourier MR imaging systems.

Detailed Description Paragraph Right (23):
FIGS. 5 and 5A illustrate the k-space matrix construction for this prototype in more detail. Half the usual number of gradient steps with twice the usual spacing in k-space were applied, and the RF response was recorded in each of three coils. Given the gradient phase encoding regimen just described, each of these RF responses corresponded to an image of half the desired field of view. The coil signals were then combined into two sets with different weightings to produce synthetic composite signals corresponding to the zero.sup.th and first harmonics, and the two synthesized responses were interleaved to fill alternating lines of the k-space matrix. The matrix was then Fourier transformed in a conventional way to form the image. The resulting image had full resolution over the full field of view. As shown in FIG. 5, line (A) depicts the acquisition of signal in each component coil. A representative signal point P is picked up in each coil, and thus contributes to the line in k-space formed with each coil signal. Line (B) shows the processing performed in the prototype described above, according to which the coil signals are combined into a first set S.sub.0 and a second set S.sub.1 which each form different lines (even and odd as shown, corresponding to one gradient step separation) of the k-space matrix. These are combined at line (C) into the full matrix.

Detailed Description Paragraph Right (24):
FIG. 5A illustrates corresponding images and signals, Line (A) of that FIGURE shows the half field-of-view images a, b, c reconstructed from separate coil sgnals, of which the signals themselves are shown in line (B). Line (C) shows the two spatial harmonic signals and line (D) the full interleaved signal, or k-space matrix. Line (E) illustrates the image reconstructed from the signals of line (C). The middle three stages (B), (C) and (D) correspond directly to the steps shown schematically in FIG. 5.

Detailed Description Paragraph Right (26):
Similar results were obtained in vivo. FIGS. 7A and 7B show a reference and SMASH-reconstructed coronal image through the brain of a healthy adult volunteer, the images being acquired in 71 and 35 seconds, respectively. All these images were acquired using commercial hardware and a convenient pulse sequence with high spatial resolution and good SNR; a fifty percent acquisition time reduction was achieved. The acquisition time savings described here have also been shown to apply to other commercial machines and pulse sequences, including some of the fastest MR imaging machines and sequences. Some residual foldover artifacts are present in the SMASH reconstructed image, due to imperfections in the composite spatial harmonic sensitivity profiles. However, the prototype used an existing coil array designed for other purposes, and improvements both in coil design and in the accuracy of coil sensitivity mapping and spatial harmonic generation, are expected to minimize these artifacts.

Detailed Description Paragraph Right (28):
The raw data for the foregoing images were generated on a commercial 1.5 Tesla whole body clinical MR imager, which was a Philips NT, made by Philips Medical Systems of Best, Netherlands. For the phantom images, a standard circular phantom of 200 mm diameter and varied internal structure was used. At certain levels, the extire extent of the phantom was uniformly filled with water, whereas at other levels it was separated into smaller compartments of varied geometry. The different regions of this phantom were used either as sensitivity references (uniform structure) or for test images (varied structure). A prototype coil array on hand in applicant's laboratory was used as the SMASH coil array. The equivalent circuit for this coil consisted of three rectangular component coils arranged in a linear array with a slight overlap in the right-left direction, similar to the array shown schematically in FIG. 2B. As these coils were designed for another purpose, details of the coil circuitry not relevant here resulted in more complicated sensitivity profiles than would be expected from the simple geometry described above. It was these sensitivity variations which were primarily responsible for the residual foldover artifacts noted in FIGS. 6B and 7B. A simpler coil designed with spatial harmonic generation in mind will circumvent these artifacts.

Detailed Description Paragraph Right (29):
The phantom test images of FIG. 6A, B were generated as follows. First with the phantom centered over the coil array, data for the reference image were acquired in a 6 mm thick coronal slice parallel to and 80 mm above the plane of the array, using

a turbo spin echo pulse sequence with five echoes per excitation. The field-of-view (FOV) was 200 mm, centered on the phantom, and matrix size was 256.times.256. Phase encoding was performed in the right-left direction (i.e. in the direction of the coil array), with a single signal average. Data from each of the three component coil channels was acquired simultaneously and, in the prototype, were stored separately for later processing. Acquisition time was measured at 10 seconds. Next, a second coronal slice using the same technique and imaging parameters was taken to serve as a measure of component coil sensitivity. It was acquired 12 mm above the first one, in a region of uniform spin density in the phantom. Then, a third image at the same level as the first was obtained in half the time using twice the phase-encode step, and hence half the field of view, in the right-left direction. Matrix size was now 256.times.128. Acquisition time was 5 seconds, exactly half the time taken for the first image.

Detailed Description Paragraph Right (30):
For the in vivo images of FIGURIE 7A, B, the volunteer was positioned with his head above the coil array, and images were taken in the same plane and with the same parameters as for the phantom images, except that eight signal averages with a slice thickness of 10 mm were used to improve SNR. The coil sensitivity image from the phantom was also used as a sensitivity reference for the in vivo images.

Detailed Description Paragraph Right (32):
Optimal weights for linear combination of component coil signals were determined by iterative fitting of this sensitivity data to the target spatial harmonic sensitivity profiles, using a gradient descent fitting routine with the complex weights serving as fitting parameters and the sum of absolute magnitude deviations from the target profile serving as a measure of goodness of fit. Two target spatial harmonics were fit in this fashion: the zero-frequency harmonic corresponding to uniform sensitivity, and the first harmonic having a modulation wavelength equal to the field-of-view of 200 mm. The results, shown in panels C and D of FIG. 4A, demonstrates that very close fits to the target harmonics may be achieved by component coil weighting, even with only three coils. A slight residual ripple was present in the zeroth harmonic fit and this is primarily responsible for imperfections in the SMASH image reconstructions.

Detailed Description Paragraph Right (33):
FIG. 5A shows intermediate stages in SMASH reconstruction of the phantom image. The procedure for the in vivo image was identical. Using weights from the iterative fit, the three component coil signals (IB) representing half-time, half-FOV aliased images (A) were combined into two composite signal sets, one for the zero.sup.th spatial harmonic and one for the first spatial harmonic (C). Finally, the two composite signal data sets were interleaved to form a data matrix of size 256.times.256 (D), This matrix was Fourier transformed to yield the reconstructed image (E), which is also shown in FIG. 6B.

Detailed Description Paragraph Right (35):
In general, the foregoing SMASH imaging technique may be applied with a great many of the known pulse sequences or spin conditioning techniques, and will in general share the advantages of the underlying sequential imaging methods which are used to collect partial k-space information. As long as suitable spatial harmonics may be generated with a coil array, the additional acquisition time savings afforded by SMASH reconstruction involves no significant sacrifice in resolution or SNR. This contrasts markedly with the trade-off in SNR or resolution that characterizes many existing approaches to fast imaging, such as low flip-angle sequences.

Detailed Description Paragraph Right (36):
Visual inspection of the phantom images in FIGS. 6A, 6B reveals a slight degradation in SNR in the SMASH reconstructed image as compared with the reference image. Part of this apparent loss may be traced to the residual foldover artifacts, in which some of the intensity of the primary image is "stolen" by aliased ghosts. There is another noteworthy difference between the SNR profiles of the two images, however. The conventional reference image in FIG. 6A was generated using a sum-of-squares combination of component coil images, as is described in PB. Roemer, W. A. Edelstein, C. E. Hayes, S. P. Souza, and O. M. Mueller, The NMR phased array. Magn. Reson. Med. 16, 192-225 (1990) and as has become a standard practice in phased array imaging to improve SNR. This combination algorithm yields an essentially constant noise profile across the image, with varying signal (and hence varying SNR) that is enhanced in regions of significant overlap between component coil sensitivities. SMASH reconstruction, on the other hand, results in a constant-signal image with a

varying noise profile. The linear combinations of component coil signals in SMASH are explicitly designed to produce a homogeneous composite signal. This is most obvious in the case of the zero.sup.th harmonic combination, which produces a flat net sensitivity profile C.sup.tot across the image plane. While the higher harmonics do involve significant spatial sensitivity variations, their profiles are all complex exponentials of unit modulus, and they do not lead to any intensity variations in the absolute magnitude image. (In other words, any apparent loss of sensitivity in the real channel is precisely compensated by a sensitivity gain in the imaginary channel.) SMASH reconstructions, therefore, do away with intensity peaks in regions of component coil overlap in favor of a spatially homogeneous image profile. Thus, sensitivity-dependent linear combinations may be used independently of SMASH as a method of homogeneity correction. A homogeneity-corrected version of the image in FIG. 6A may be generated by linear combination of component coil reference images, using the weights calculated for the zero.sup.th harmonic profile in FIG. 4A. Roemer et. al. have described an alternative combination agorithm for producing constant-signal images from phased arrays, and SMASH images produced with accurate spatial harmonics will have similar intensity profiles.

Detailed Description Paragraph Right (40):
Finally, the geometry of the coil array will place certain limitations on the field of view, the position, and the angulation of image planes suitable for SMASH reconstruction. Surface coil sensitivity profiles vary with distance from the coil center, and while simulations indicate that a wide range of image plane geometries are compatible with SMASH reconstruction, the reconstruction may begin to fail at large distances and angles, where sensitivity functions become broad and asymmetric. These constraints will be relaxed in the presence of larger numbers of component coils whose sensitivities provide good coverage of the imaged volume. The design of RF coil arrays specifically tailored for accurate and flexible spatial harmonic generation will help to overcome the limitations outlined above.

Detailed Description Paragraph Right (41):
In particular, the invention contemplates the use of coil arrays with multiple component coils extending in more than one linear direction. A two-dimensional array, such as an N xM rectangular array, will allow generation of spatial harmonics along multiple directions and will relax constraints on image plane position and angulation. Wraparound arrays are also contemplated to allow spatial harmonic generation in a plane transverse to the body plane. A coil arrangement with a surface array on the top of the body and another surface array on the bottom, possibly with some linear offset with respect to one another, will allow fine tuning of spatial harmonics in a plane between the two and will have the added advantage of increasing overall SNR in such a plane. An extended grid coil may also be designed with a sensitivity profile more closely approximating a sinusoid. All such multiple-coil configurations will allow improvement in the accuracy of spatial harmonic generation, as well as allowing a greater number and variety of harmonics to be generated. As in the prototype embodiment described, the component coils in these arrays may be overlapped and may be output to low input impedance preamplifiers as necessary, to minimize inductive coupling. It is also possible that the simple expedient of transforming the coil sensitivities to achieve the spatial harmonic reconstruction procedure of applicant's SMASH invention may be applied in large arrays of coils of the sort proposed by Hutchinson and Raff in 1988 and described further by Kwiat, Einav and Navon, supra.

Detailed Description Paragraph Right (42):
It should be observed that, although recent studies have shown that the sensitivity advantages of phased arrays peak at coil numbers less than 10, the advantages of faster imaging with SMASH imaging may be further enhanced by arrays with larger numbers of coils and with new coil geometries. As one may intuit from FIGS. 3 and 4A, the higher the frequency of the spatial harmonic desired, the larger the number of component coils that will be required for its generation. This is in part because, at least for simple component coil shapes, the sharpest features of the net sensitivity function can only be on the order of the component coil dimension. As noted above, the extra sensitivity information from additional coils can also be used for the fine tuning of spatial harmonics and the improvement of image quality.

Detailed Description Paragraph Right (43):
The SMASH technique as applied to a conventional NMR apparatus described above partially replaces gradient phase encoding by a spatial encoding procedure tied to the detection coils, in which some of the spatial modulations that distinguish different phase-encoding lines are generated by combining signals from multiple

coils arrayed above or around the imaging volume. This shift of responsibility from gradient geometry to coil geometry, and from the spin preparation stage to the stage of signal detection and combination, allows for simultaneous acquisition of multiple lines of k-space. Enhancement of image speed, resolution or field of view by a factor of two, five, ten or more may be expected with appropriate coil arrays generating sufficient harmonics.

Detailed Description Paragraph Right (44):
Apart from its promise for progressively faster acquisitions, SMASH has a number of practical advantages as a fast imaging scheme. As a partially parallel acquisition strategy, it can be combined with most existing sequential fast imaging techniques for multiplicative time savings. No special hardware is required, other than an Appropriate coil array.

Detailed Description Paragraph Right (45):
One embodiment of the invention includes a coil array together with a digital signal processor configured to combine the outputs of the component coils with appropriate weights as described above, and produce two or more output signals, each of which represents a composite spatial harmonic as described above. The array with processor may then be directly substituted for a conventional receiving coil, with the difference that it produces two or more lines of data for each spin conditioning cycle of the MRI apparatus. As such, the plug-in coil unit may operate with older MRI devices to directly enhance the acquisition time, field of view, or resolution by a factor of two or more, without changing the expensive magnet and other spin conditioning hardware of the device. Commercially available or home-built MR phased arrays may well suffice to yield significant time savings, and SMASH may be performed on machines not equipped with much more costly magnetic enhancements such as EPI gradient systems. Combination of the component coil signals may be performed after the fact, allowing for a wide range of postprocessing steps, including fine tuning of spatial harmonics, adaptive artifact correction, or even non-Fourier (e.g. wavelet) encoding and reconstruction. Furthermore, it should be noted that the invention further contemplates that both for constructing the basic component coil transforms or weights and for any tuning or adaptive adjustments, the invention may include a digital signal processor or neural network processor in addition to the normal complement of control and processing assemblies in conventional apparatus to evaluate the responses or coil signals and carry out the signal combination into spatial harmonics in real time. Even in the simplest embodiments, spatial harmonic reconstructions automatically yield a homogeneous intensity profile for the reconstructed image, and this may be advantageous for some imaging applications.

Detailed Description Paragraph Right (47):
The invention has been described above with reference to a prototype embodiment which produces signals corresponding to spatial harmonics that yield additional k-space lines equivalent to the conventional gradient steps in a Fourier transform imaging system and a simple coil array and processing assembly for enhancing image speed, quality or field. However, the method is broadly applicable to coils and processors for producing various spatial representations, including wavelets or other non-Fourier spatial representations to replace, supplement or enhance MR imaging processes and systems. The invention being thus described, further variations and modifications will occur to those skilled in the art, and all such variations and modifications are considered to be within the scope of the invention and its equivalents, as set forth in the claims appended hereto.

Other Reference Publication (3):
Kwiat, D. and Einav, S., "A Decoupled Coil Detector Array for Fast Image Acquisition in Magnetic Resonance Imaging," Med. Phys., vol. 18, No. 2, 251-265 (1991).

Other Reference Publication (4):
Kwiat, D. and Einav, S., "Preliminary Experimental Evaluation of an Inverse Source Imaging Procedure Using a Decoupled Coil Detector Array in Magnetic Resonance Imaging," Med. Eng. Phys., vol. 17, No. 4, 257-263 (1995).

CLAIMS:

1. A method of magnetic resonance imaging of a continuous region of a body by conditioning nuclear spins and measuring RF signals indicative of the conditioned spins, wherein the method includes performing multiple steps of spin conditioning and of collecting of RF measurement signal responses from the region, establishing an ordered data set of collected RF signals, and performing a spatial transformation

of the ordered data set to produce a magnetic resonance image of said region, wherein the method includes the steps of

1) for a spin conditioning step

i) receiving an RF measurement signal response simultaneously in each of plural RF receiving coils which each have a different spatial sensitivity, thereby acquiring a plurality of response signals

ii) forming at least two distinct combinations of said plurality of response signals, including a first combination signal and one or more remaining combination signals

iii) applying said first combination signal to an entry of said ordered data set that corresponds to said spin conditioning step, and applying the remaining combination signals to fill further entries of said ordered data set which are offset from said entry that corresponds to the spin conditioning step,

2) performing additional spin conditioning steps and repeating steps i)-iii) for each additional spin conditioning step to thereby fill said ordered data set while reducing spin conditioning steps performed, and

3) performing a spatial transformation of the filled and ordered data set into a magnetic resonance image of said continuous region thereby imaging said continuous region with reduced data acquisition time.

2. A method according to claim 1, wherein the spin conditioning steps include steps of applvini at least one of RF pulses and magnetic field gradients and said steps of receiving and combining are carried out to form images of a given field of view and spatial resolution while reducing at least one of number or duration of spin conditioning steps performed and thereby reducing image acquisition time.

12. A method of imaging a region of a body by MR by capturing MRI data during multiple cycles of NMR spin conditioning and measurement, said method comprising the steps of:

i) during one spin conditioning cycle receiving an NMR RF measurement signal response simultaneously in each of plural surface coils having different spatial sensitivities,

ii) combining said NMR RF measurement signal responses received in the plural coils by linear combinations with complex coefficients into plural sets of composite NMR signals, each set of composite NMR signals containing different spatial information wherein the step of combining signal responses yields composite spatial sensitivities which approximate spatial harmonics of an imaged field of views.

iii) performing additional spin conditioning cycles and repeating said steps of receiving and combining to form a data set augmented by at least one set of said composite NMR signals, and

iv) forming an MR image from said data set.

14. A method according to claim 12 wherein the step of receiving in surface coils includes receiving in at least one of a two dimensional coil array, an N.times.M coil array, a wrap around coil array and an extended grid coil array.

15. A method according to claim 12, further comprising the step of mapping sensitivity of said plural coils to determine weights for combining the received NMR RF measurement signal responses into said sets of spatial harmonic composite NMR signals.

17. A magnetic resonance imaging (MRI) apparatus of the type having a magnet for establishing a magnetic field, means for conditioning nuclear spins in a continuous region and means for collecting a signal response from the conditioned spins, said means for conditioning and said means for collecting being operated in repetitive cycles to collect signal responses in an ordered data set, and including a processor which applies a spatial transformation to the ordered data set to produce a magnetic resonance image having a characteristic resolution and field of view over said region, wherein the apparatus includes

a plurality of coils for receiving said signal responses, the coils forming an array and each having a different spatial sensitivity so as to each receive a different signal response, and

means for combining the signal responses received in one conditioning step in said plural individual coils of said array to form a plurality of distinct combination signals corresponding to independent spatial representations of spin data from said region, each of said plurality of distinct combination signals being applied to fill an entry of said ordered data set which is offset from other entries of said data set, and thereby filling said data set with multiple combination signals formed from the signal responses received in each conditioning step

said processor applying a spatial transform to the filled ordered data set to thereby produce said image having the characteristic resolution and field of view over the region while reducing required spin conditioning.

23. A magnetic resonance imaging (MRI) apparatus according to claim 17, wherein said plurality of coils are surface coils which form an array having an extent of approximately twenty to about fifty centimeters, and individual coils of the array each have an extent under about twenty centimeters.

24. A magnetic resonance imaging (MRI) apparatus according to claim 17, wherein said plurality of coils form a linear array.

25. A magnetic resonance imaging (MRI) apparatus of the type having a magnet for establishing a magnetic field in a region, means for conditioning nuclear spins in the region and means for collecting an RF signal from the conditioned spins, said magnet and said means being operated in repetitive cycles to collect spin data and construct an image of material in said region, wherein said means for collecting an RF signal includes

a plurality of surface coils forming an array, and each having a different spatial sensitivity, and

means for combining signals from plural individual coils of said array with different weights into a plurality of sets of signals corresponding to orthogonal spatial representations of spin data from said region, wherein said spatial representations are spatial harmonics.

26. In a magnetic resonance imaging (MRI) apparatus of the type having a magnet for establishing a magnetic field in a continuous region, means for conditioning nuclear spins in the region and means for collecting a magnetic resonance response signal from the conditioned spins, said means for conditioning and said means for collecting being operated in repetitive cycles to collect response signals and construct a spatially ordered data set of signal data which is then transformed to form an image of material in said region, the improvement wherein said means for collecting a magnetic resonance response signal includes a plurality of surface coils forming an array, and each having a different spatial sensitivity, and the apparatus further includes

means for combining signals from plural individual coils of said array with different weights so as to form a plurality of combination signals each corresponding to an independent spatial representation of spin data from said continuous region, and said apparatus applies said combination signals to fill offset entries of said spatially ordered data set to complete the data set with reduced spin conditioning and form said image.

L9: Entry 3 of 4 File: USPT Oct 21, 1997

DOCUMENT-IDENTIFIER: US 5680047 A
TITLE: Multipl-tuned radio frequency coil for simultaneous magnetic resonance imaging and spectroscopy

Abstract Paragraph Left (1):
A magnetic resonance system is used to excite high .gamma. dipoles, such as hydrogen, and lower .gamma. dipoles, such as phosphorous, to resonate concurrently. A multiply-tuned radio frequency coil (40) is disposed around the region of interest. The multiply-tuned radio frequency coil is tuned to the resonance frequency of the high .gamma. dipoles and the resonance frequency of the low .gamma. dipoles. The coil has an inner coil section, defined by a first leg or ring (92) and a second leg or ring (94), which is tuned by added capacitance substantially to the resonance frequency of the low .gamma. dipole. A first outer coil section which is defined by a third leg or ring (90) and the first leg or ring. A second outer coil section defined by the second leg or ring (94) and a fourth leg or ring (96). The first and second outer sections, together with the inner coil section define a coil which has a co-rotating mode and a counter-rotating mode. Added capacitance is sized such that one of the co-rotating and counter-rotating modes is at the high .gamma. resonance frequencies. In this manner, a multiply-tuned radio frequency coil is defined which is simultaneously tuned to at least two resonance frequencies. Separate pick-ups (130, 132) for the high and low resonance frequencies enable both signals to be received (66, 70) simultaneously for reconstruction (74) into image representations.

Brief Summary Paragraph Right (4):
Another doubly-tuned coil design which utilizes four parallel wind solenoid coils is described in "Doubly-Tuned Solenoidal Resonators For Small Animal Imaging and Spectroscopy At 1.5 Tesla", Ballon, et al., Mag. Res. Imaging, 7, 155-162 (1989) and "Doubly-Tuned Solenoidal Resonators For High Spatial Resolution Small Animal Imaging and Spectroscopy", Ballon, et al., SMRM 7th Annual Meeting, Book of Abstracts, p. 861 (1988). This solenoidal coil design had several resonant modes including (1) a parallel mode in which the currents in all four turns were in-phase and (2) an anti-parallel mode in which the currents in the two outer turns were out-of-phase with respect to the currents in the two inner turns. The parallel mode, which has the larger field-of-view was at the lower resonance frequency and the anti-parallel mode with the narrower field-of-view was at the higher resonance frequency. Generally, the proton mode with a larger field-of-view than the low .gamma. nucleus is preferred for (1) obtaining a homogeneous image, (2) shimming over a larger volume, and (3) uniform proton decoupling over the proton mode field-of-view. These all promote obtaining quality x-nucleus spectra. Ballon added additional external tuning circuits in order to cause the parallel mode to resonate at the higher hydrogen frequency and the anti-parallel mode to resonate at the lower phosphorous frequency. In this manner, added tuning traps were needed to produce both resonance frequency modes.

Brief Summary Paragraph Right (5):
In the Ballon design, the hydrogen dipole field was relatively homogeneous. The hydrogen dipole field was relatively homogeneous due to a large fraction of the currents flowing in the inner turns. However, the strength of the B.sub.1 radio frequency field at the phosphorous frequency was reduced appreciably at the center by the opposite polarity contribution of the two outer rings. The .sup.31 P radio frequency field was reduced at the center due to the equal but 180.degree. out-of-phase currents in the outer turns. This not only caused a reduction in the phosphorous B.sub.1 radio frequency field at coil center, but also forced nulls between each of the inner turns and its nearest neighbor outer turns. These nulls

caused rapidly decreasing .sup.31 P resonance frequency fields in the sample region that was disposed between the inner and their nearest neighbor outer turns. That is, little or no phosphorous resonance signals were received from tissues disposed in the examination region in the slice or slab between each pair of inner and outer turns. These nulls thus cause an unacceptable dirth of .sup.31 P resonance signals over portions of the field-of-view.

Brief Summary Paragraph Right (8):
In accordance one aspect of the present invention, a magnetic resonance apparatus is provided. A magnet generates a temporally constant uniform magnetic field through an examination. At least one multiply-tuned radio frequency coil transmits radio frequency signals into the examination region to induce and manipulate resonance of first, higher .gamma. dipoles and to receive first resonance frequency signals from the first dipoles and second resonance frequency signals from the second dipoles. A processor processes the received magnetic resonance signals. The radio frequency coil includes a first coil portion that has first and second legs disposed in parallel and electrically interconnected. The first coil portion has an inherent inductance and added capacitance such that it is tuned near the first resonance frequency. A third leg disposed in parallel to the first leg and electrically interconnected with the first leg defines a second coil portion. A fourth leg disposed parallel to the second leg and electrically connected with the second leg defines a third coil portion. The third and fourth legs have added capacitance such that the first, second, third and fourth legs taken together resonate in at least of a co-rotating and a counter-rotating mode near the second resonance frequency.

Brief Summary Paragraph Right (11):
In accordance with another aspect of the present invention, a multiply-tuned radio frequency coil is tuned to receive radio frequency magnetic resonance signals of at least a first resonance frequency and a second resonance frequency simultaneously. An inner coil section is tuned with its inherent inductance and added first capacitive elements substantially to the first resonance frequency. A first outer coil section and a second outer coil section are connected with opposite sides of the inner coil section. The first and second outer coil sections have added second capacitive elements. The second capacitive elements are sized relative to the inherent inductance of the inner coil section and the first and second outer coil sections, the first capacitive elements, and the coil size geometry to tune the coil defined by the inner and the first and second outer coil sections operating together in one of a co-rotating and a counter-rotating mode to the second resonance frequency. The second capacitive elements are sized such that at the first resonance frequency, radio frequency current flow through the inner coil section is much larger than the radio frequency current flow through the outer coil sections. In this manner, the multiply-tuned radio frequency coil is tuned to the first and second resonance frequencies without added traps.

Brief Summary Paragraph Right (12):
In accordance with a more limited aspect of the present invention, the inner coil section is defined by first and second legs and an electrical connection therebetween. The first outer coil section is defined by a third leg, the first leg, and an electrical interconnection therebetween. The second outer coil section is defined by a fourth leg, the second leg, and an electrical interconnection between the third and fourth legs.

Brief Summary Paragraph Right (13):
In accordance with another aspect of the present invention, the first, second, third, and fourth legs are loops or rings such that the multiply-tuned radio frequency coil is solenoidal.

Brief Summary Paragraph Right (14):
In accordance with another more limited aspect of the present invention, the first, second, third, and fourth legs are linear such that the multiply-tuned radio frequency coil is planar.

Brief Summary Paragraph Right (15):
In accordance with another aspect of the present invention, a method of magnetic resonance imaging is provided. A temporally constant uniform magnetic field is generated through an examination region. Radio frequency signals are transmitted into the examination region to induce and manipulate magnetic resonance of first dipoles which resonate at a first frequency and second dipoles which resonate at a second frequency. First resonance signals and second resonance signals are received

from the resonating first and second dipoles in the examination region. Magnetic field gradients are applied across the examination region. The received first and second frequency signals are processed into image representations. The first and second resonance frequency signals are received simultaneously with a multiply-tuned radio frequency coil which is free of added trap circuitry.

Brief Summary Paragraph Right (21):
In accordance with another aspect of the present invention, additional degrees of design freedom including varying coil geometry (by varying dimensions, impedance, and the like) permits optimization of the signal-to-noise ratio and the B.sub.1 profile.

Brief Summary Paragraph Right (24):
Another advantage of the present invention is that a high frequency co-rotating or counter-rotating mode provides a uniform B.sub.1 radio frequency field over and beyond the low frequency mode field-of-view.

Brief Summary Paragraph Right (25):
Another advantage of the present invention resides in additional degrees of freedom in coil design for optimizing the signal-to-noise ratio, the B.sub.1 field profile, and the like.

Detailed Description Paragraph Right (2):
A whole body gradient coil assembly 30 includes x, y, and z-coils mounted along the bore 12 for generating gradient magnetic fields, G.sub.x, G.sub.y, and G.sub.z. Preferably, the gradient coil assembly is a self-shielded gradient coil that includes primary x, y, and z-coil assemblies 32 potted in a dielectric former and secondary x, y, and z-coil assemblies 34 that are supported on a bore defining cylinder of the vacuum dewar 20. A whole body radio frequency coil 36 is mounted inside the gradient coil assembly 30. A whole body radio frequency shield 38, e.g., copper mesh, is mounted between the whole body RF coil 36 and the gradient coil assembly 30.

Detailed Description Paragraph Right (5):
The higher frequency, .sup.1 H resonance frequency signals are demodulated by a first digital receiver 66 and stored in a data memory 68. The lower frequency resonance signals, such as the .sup.31 P resonance signals are received by a second receiver 68 and stored in a second data memory 72. Of course, the first and second data memories may be portions of a large mass memory. Data from the data memories are reconstructed by a reconstruction or array processor 74 into corresponding volumetric image representations that are stored in corresponding portions of an image memory 76. Again, the image memory may be part of a common mass memory. The two volume images or selected portions of them are adjustably weighted and combined by a video processor 78 under operator control. The selected, relatively weighted portions of the images, such as slice images, projection images, perspective views, or the like, as is conventional in the art, are displayed on the video monitor 52.

Detailed Description Paragraph Right (10):
With reference to FIGS. 5A and 5B, in addition to the co-rotating and .sup.31 P modes, there is also a counter-rotating mode. In the counter-rotating mode, the currents in the two outer meshes are 180.degree. out-of-phase with one another with substantially no net current flowing between the two inner rings. This provides a linear gradient along the coil and produces no net field at coil center. This gradient RF field is usable for rotating frame experiments. See Hoult, J. Mag. Res., 33, p. 183 (1979). Optionally, the counter-rotating mode can be tuned to the hydrogen or phosphorous frequencies.

Detailed Description Paragraph Right (16):
The above-described coils can be overlapped, preferably to a point of minimum mutual inductance with like coils or other surface volume coils. Such overlapped coils may be used individually or collectively, depending on the application, as is known in the art. These coils may further be overlapped with other coils, including surface coils, volume coils, single-tuned coils, multiply-tuned coils, or the like, in an array configuration. The coils may be used for imaging of both dipoles or for imaging of one and spectroscopic examination of the other. With separate pick-ups for the high and low .gamma. signals, the high and low .gamma. magnetic resonance frequency signals are preferably generated simultaneously, yet received independently. The coil may be used for receive-only, transmit-only, or transmit and receive purposes. The coil may also be used either alone or in conjunction with

local gradient coils for high resolution or rapid imaging.

CLAIMS:

1. In a magnetic resonance apparatus which includes a magnet for generating a temporally constant uniform magnetic field through an examination region, at least one multiply-tuned radio frequency coil which performs at least one of (1) transmitting radio frequency signals into the examination region to induce and manipulate resonance of first, higher .gamma. dipoles and second, lower .gamma. dipoles disposed in the examination region, and (2) receiving first resonance frequency signals from the first dipoles and second resonance frequency signals from the second dipoles, and a processor for processing the received magnetic resonance signals, the at least one multiply-tuned radio frequency coil comprising:

a first coil portion having first and second legs disposed in parallel and electrically interconnected to define the first coil portion, the first coil portion having an inherent inductance and added capacitance such that it is tuned near the first resonance frequency of the low .gamma. dipoles;

a third leg disposed parallel to the first leg and electrically interconnected with the first leg to define a second coil portion;

a fourth leg disposed parallel to the second leg and electrically connected with the second leg defines a third coil portion, the third and fourth legs having added capacitance such that the first, second, third, and fourth legs, taken together, resonate in at least one of a co-rotating and a counter-rotating mode near the second resonance frequency of the high .gamma. dipoles.

3. In the magnetic resonance apparatus as set forth in claim 1, wherein the first, second, third, and fourth legs extend around a cylindrical region such that the multiply-tuned radio frequency coil is generally solenoidal.

4. In the magnetic resonance apparatus as set forth in claim 3, the multiply-tuned radio frequency coil further including fourth and fifth legs disposed parallel to and interconnecting the first and second legs such that the low .gamma. frequency coil has at least four legs.

5. A multiply-tuned radio frequency coil tuned to receive radio frequency magnetic resonance signals of at least a first resonance frequency and a second resonance frequency simultaneously, the multiply-tuned radio frequency coil comprising:

an inner coil section tuned with its inherent inductance and added first capacitive elements substantially to the first resonance frequency;

a first outer coil section and a second outer coil section connected with opposite sides of the inner coil section, the first and second outer coil sections having added second capacitive elements therein, which second capacitive elements are sized relative to inherent inductance in the inner coil section and the first and second outer coil sections, the first capacitive elements in the inner coil section, and coil size and geometry to tune a coil defined by the inner and the first and second outer coil sections operating together in one of a co-rotating and counter-rotating mode to the second resonance frequency, the second capacitive elements in the first and second outer coil sections further being sized such that at the first resonance frequency, radio frequency current flow through the inner coil section is much larger than radio frequency current flow through the first and second outer coil sections, whereby the multiply-tuned radio frequency coil is tuned to the first and second resonance frequencies without added traps.

6. The multiply-tuned radio frequency coil as set forth in claim 5 wherein

the inner coil section includes first and second annular loops electrically connected in series; and,

the first outer coil section is defined by a third annular loop which is electrically connected electrically in parallel with the first annular loop and the second outer coil section is defined by a fourth annular loop that is connected electrically in parallel with the second annular loop.

7. The multiply-tuned radio frequency coil as set forth in claim 6 wherein

capacitive elements are disposed only in the third and fourth annular loops and in an electrical interconnection between the first and second annular loops.

8. The multiply-tuned radio frequency coil as set forth in claim 6 wherein the first and second annular loops are electrically connected by at least one additional annular loop disposed between the first and second annular loops.

9. The multiply-tuned radio frequency coil as set forth in claim 6 wherein:

the inner coil section is defined by a first leg, a second leg, and electrical elements interconnecting the first and second legs;

the first outer coil section is defined by a third leg, the first leg, and an electrical interconnection between the first and second legs; and,

the second outer coil section is defined by a fourth leg, the second leg, and an electrical interconnection between the second and fourth legs.

10. The multiply-tuned radio frequency coil as set forth in claim 9 further including capacitive elements connected in the third and fourth legs and the electrical circuitry interconnecting the first and second legs.

12. The multiply-tuned radio frequency coil as set forth in claim 10 wherein the first, second, third, and fourth legs lie parallel to each other in a common plane.

13. The multiply-tuned radio frequency coil as set forth in claim 10 wherein the first, second, third, and fourth legs are disposed parallel to each other circumferentially around a cylinder.

14. The multiply-tuned radio frequency coil as set forth in claim 10 wherein the first, second, third, and fourth legs extend along a surface of a cylinder parallel to each other and parallel to a central longitudinal axis of the cylinder.

15. The multiply-tuned resonance coil as set forth in claim 10 further including additional legs disposed between the first and second legs.

16. The multiply-tuned radio frequency coil as set forth in claim 5 wherein the other of the co-rotating and counter-rotating modes is tuned to a third frequency.

17. A multiply-tuned radio frequency magnetic resonance imaging coil comprising:

a first coil portion tuned in a first mode to a first radio frequency and having a substantially uniform B.sub.1 field over a first field-of-view;

a second coil portion connected with the first coil portion and being tuned in a second mode to a second radio frequency, the second frequency being different from the first frequency, one of the first and second modes being a counter-rotating mode and the other being a co-rotating mode, at the second radio frequency, the coil having a substantially uniform B.sub.1 field over a second field-of-view which second field-of-view is larger than the first field-of-view to improve image homogeneity, shimming over a larger region, and uniform decoupling over the second field-of-view.

18. The magnetic resonance imaging coil as set forth in claim 17 further including a third mode in which a radio frequency gradient provides a rotating frame of reference.

19. In a method of magnetic resonance imaging in which a temporally constant uniform magnetic field is generated through an examination region, radio frequency signals are transmitted into the examination region to induce and manipulate magnetic resonance of first dipoles which resonate at a first frequency and second dipoles which resonate with a second frequency, magnetic field gradients are applied across the examination region, and first resonance frequency signals and second resonance frequency signals are received from the resonating first and second dipoles in the examination region, and processed into image representations, the improvement comprising:

receiving the first and second resonance frequency signals simultaneously with a multiply-tuned radio frequency coil which is free of added trap circuitry, the

multiply-tuned radio frequency coil having (i) a low frequency mode at the first resonance frequency, (ii) a co-rotating mode, and (iii) a counter-rotating mode, one of the co-rotating and counter-rotating modes being at the second resonance frequency.

21. In the method as set forth in claim 19, the improvement further comprising:

in the low frequency mode, first magnetic resonance frequency signals circulating through an inner loop of the multiply-tuned coil, which inner loop is tuned substantially to the first resonance frequency with only much smaller magnitudes of first resonance frequency currents flowing through a pair of outer coils; and,

substantially equal magnitudes of current of the second resonance frequency flowing through the inner loop and outer loops such that the first resonance frequency has a field-of-view commensurate with the inner coil and at the second resonance frequency the multiply-tuned coil has a field-of-view commensurate with the inner and outer loops taken together.

22. In a method of magnetic resonance imaging in which a temporally constant uniform magnetic field is generated through an examination region, radio frequency signals are transmitted into the examination region to induce and manipulate magnetic resonance of first dipoles which resonate at a first frequency and second dipoles which resonate with a second frequency, first resonance frequency signals and second resonance frequency signals are received from the resonating first and second dipoles in the examination region, magnetic field gradients are applied across the examination region, and the received first and second frequency signals are processed into image representations, the improvement comprising:

tuning a multiply-tuned radio frequency coil which is free of added trap circuitry to (1) a first mode at a first resonance frequency, (2) a second mode at the second resonance frequency, and (3) a third mode at a third frequency, one of the modes being a co-rotating mode and one of the modes being a counter-rotating mode;

receiving the first and second resonance frequency signals simultaneously with the multiply-tuned radio frequency coil.

End of Result Set

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L9: Entry 4 of 4 File: USPT Feb 4, 1992

DOCUMENT-IDENTIFIER: US 5086275 A
TITLE: Time domain filtering for NMR phased array imaging

Abstract Paragraph Left (1):
A method and apparatus for combining NMR response data of a sample from a plurality of closely spaced RF receiver coils of an NMR phased array in the time domain to form a composite NMR image wherein each of the RF receiver coils receives a different respective one of a plurality of NMR response signals, each of which is evoked from a portion of the sample within a field of a respective one of the receiver coils. The response signals are conditioned to develop a plurality of data point signals corresponding to the magnitude of each of the respective response signals from each of the receiver coils at successive time intervals. The data point signals are convolved by a time domain representation of a field map of the respective one of the receiver coils generating the corresponding one of the response signals. The convolved signals are combined on a time domain point-by-point basis to produce a time domain representation of the composite NMR image of the sample.

Brief Summary Paragraph Right (1):
The present invention relates to nuclear magnetic resonance (NMR) imaging and, more particularly, to methods and apparatus for combining the simultaneously received data from a plurality of radio-frequency (RF) coils of an NMR phased array in the time, rather than image, domain to produce a composite image having high signal-to-noise ratio (SNR) throughout the image.

Brief Summary Paragraph Right (2):
The term "NMR phased array" refers to apparatus, such as shown in Roemer et al. U.S. Pat. No. 4,871,969 (the disclosure of which is incorporated herein by reference), wherein a plurality of closely-spaced RF coils is employed for simultaneously receiving different NMR response signals from associated portions of a sample (such as a patient in medical imaging) and combining the separate data from each coil to produce a single composite NMR image of the sample. By overlapping adjacent coils and connecting each coil to the input of an associated low-input-impedance preamplifier channel, the high SNR of a single surface coil can be maintained over fields-of-view (FOV) characteristic of remote coils.

Brief Summary Paragraph Right (3):
Currently, composite images for NMR phased arrays are reconstructed in the image domain by combining the individual image contributions on a weighted, point-by-point basis after first acquiring the complete NMR images for each separate coil. The reason for acquiring the separate images first is that the optimum set of weights needed to maximize SNR when combining the separate signals to produce the composite image is a function of position, and so varies from point to point. While the phase shifters and transformers of the setup shown in FIG. 6 of Roemer et al. U.S. Pat. No. 4,825,162 can be adjusted to provide a composite image in the time domain having a high SNR at any particular point, different weighting must be applied for each point in order to obtain good sensitivity over the whole image. Thus, the conventional approach is to first separately acquire the different NMR image from each coil before combining the different individual images, on a point-by-point basis, to form the composite image.

Brief Summary Paragraph Right (4):
NMR phased array imaging as described in the '162 patent, therefore, has the drawbacks of requiring large amounts of memory to store the separate coil images

before reconstruction and of necessitating long time delays between acquisition of the last data point and onset of the first display of the reconstructed image.

Brief Summary Paragraph Right (5):
It is desirable in NMR phased array imaging to be able to combine the data from the separate receiver coils as it is acquired on a time domain, rather than image domain, basis without sacrificing SNR resolution. Combining the data as acquired will reduce the total memory requirements of the system since only one combined data set would have to be stored and, because only the combined data set will have to be transformed at the end of scanning, will also reduce the time between end-of-scan and first appearance of the composite image.

Brief Summary Paragraph Right (7):
In accordance with the invention, a method is provided for combining the simultaneously received different NMR response signals from a plurality of closely-spaced, overlapping RF receiver coils of an NMR phased array in the time domain, to form a composite image that has high SNR throughout the image. A filter scheme is utilized to develop a composite data set in the time domain, wherein each time point of the composite data is formed on the basis of contributions from previous data points and future data points. The data is passed through filter arrangements having one-, two- and three-dimensional filters before the signals are summed together. Each filter dimension corresponds to filtering in one of the time dimensions of k-space, i.e., the readout direction; the phase encode direction; and, in the case of three-dimensional imaging, the second phase encode direction. Filter coefficients are chosen to combine the data in a way that is simultaneously optimal for providing a high SNR at multiple points of the composite image. With more terms added to the filter, the SNR can be optimized over the entire image.

Brief Summary Paragraph Right (8):
In a preferred embodiment, described in greater detail below, the filter functions are determined from the RF magnetic field profiles of the receiving coils.

Drawing Description Paragraph Right (1):
FIG. 1 (prior art) is a schematic view of an arrangement employed in the conventional method to combine the signals from the overlapping coils of an NMR phased array using image domain data processing techniques.

Drawing Description Paragraph Right (5):
FIG. 5 is a schematic view of an arrangement employed in a method of combining the coil signals of an NMR phased array in the time domain using filters in accordance with the present invention.

Detailed Description Paragraph Right (1):
FIGS. 1 and 5 show an NMR phased array 10, such as described in Roemer et al. U.S. Pat. No. 4,825,162, of a plurality of radio-frequency (RF) receiver coils 12 (coils 1 through N.sub.c) defining an imaging volume for the NMR imaging of a sample, such as for the NMR medical diagnostic imaging of a human spine. The separate surface coils 12 are identically configured and are arranged in closely-spaced relationship with overlapping fields-of-view (FOV), but with substantially no interaction between adjacent coils. The coils 12 are adapted as part of the NMR imaging process to simultaneously receive a different one of a plurality of NMR response signals each evoked from an associated portion of the sample enclosed in the imaging volume. As shown, each coil 12 has its own processing channel 14 including receiver circuitry 15 and an analog-to-digital converter 16. FIG. 1 is a schematic representation of the conventional data processing set-up for constructing a different NMR image for each channel 14 of a sample portion from the NMR response signals received by the associated coil 12 for that channel 14, and for subsequently combining the plurality of different images thus constructed, on a point-by-point basis, in the image domain, to produce a single final NMR image of all sample portions from which an NMR signal was received by any of the coils 12. FIG. 5 is a schematic representation of the corresponding set-up for performing the image reconstruction in the time domain utilizing the principles of the present invention.

Detailed Description Paragraph Right (2):
As described in Roemer et al. U.S. Pat. No. 4,871,969, the optimal combination or weighting of signals from the individual coils 1-N.sub.c in the array 10 to achieve a high signal-to-noise ratio (SNR) is dependent on the location (x, y, z) of a particular volume element (voxel). This is because the signal of each RF receiving coil C.sub.i is sensitive to nuclear spins in proportion to the field B.sub.i

created by the coil, whereas the noise is "white noise" uniformly distributed over the image. Hence, the resultant SNR is a function of position.

Detailed Description Paragraph Right (3):
Assume that I.sub.1 (x,y) is the complex image obtained by reconstructing the data received from coil C.sub.i, and B.sub.i (x,y) is the RF magnetic field produced by coil C.sub.i. The real part of B is the x component (in magnet coordinates as opposed to the screen coordinates of the image) of the transverse RF magnetic field and the imaginary part of B is the y component of the field. If noise correlations are ignored (which will have little effect on image quality) and all coils 12 have approximately the same noise, the combination of separate images I.sub.i that optimizes the SNR in the composite image is given by ##EQU1## where I(x,y) is the composite image.

Detailed Description Paragraph Right (4):
The complex image is really the product of the RF receiving coil magnetic field and the spin density S(x,y) given by

Detailed Description Paragraph Right (8):
To derive the time domain filtering method of the present invention (FIG. 5), it was recognized that the combined image obtained using the image domain method is simply the Fourier transform of the original time dependent data. In accordance with equation (1), the optimal combination of images I(x,y) (i.e., that giving high SNR over the whole image) is obtained by multiplying each separate coil image I.sub.i (x,y) by its corresponding RF coil magnetic field map profile B.sub.i (x,y) at 21, before summing the results at 20 (see FIG. 1). From linear system theory, however, it is known that convolution in the time domain is equivalent to multiplication in the spatial domain. Thus, for a single slice of multi-slice data the time domain representation of the composite image can be given by the two-dimensional convolution integral ##EQU3## where N.sub.c is the number of coils, v.sub.i (t.sub.r,t.sub..PHI.) is the time dependent NMR voltage signal measured on coil i, b.sub.i (.tau..sub.r,.tau..sub..PHI.) is the inverse Fourier transform of coil i's RF field map, A(t.sub.r,t.sub..PHI.) is the inverse Fourier transform of the composite data set and t.sub.r is the readout time for each phase encode time t.sub..PHI..

Detailed Description Paragraph Right (9):
For a finite set of discrete samples, the inverse Fourier transform A(j,k) of the composite image is ##EQU4## where v.sub.i (j,k) is a matrix of NMR voltages measured on coil i and b.sub.i (j,k) is the discrete Fourier transform of the field map from coil i. The first and second arguments are the sample indices in the readout and phase encode directions, respectively, and N.sub.r and N.sub..PHI. are the number of samples in the readout and phase encode directions, respectively.

Detailed Description Paragraph Right (11):
The number of computations required for the convolution can be greatly reduced, however, through the recognition that the RF field map 21 (shown in FIG. 1) are relatively slowly varying quantities across the image and can thus be suitably represented in abbreviated form for time domain processing purposes. It has been observed that the inverse Fourier transform of the field map is concentrated near the origin in the time domain (k-space) and thus the b.sub.i (1,m) terms in equation (5) can be truncated to a kernel containing relatively few terms.

Detailed Description Paragraph Right (12):
By way of example, FIG. 2 shows the magnitude of a calculated sensitivity profile of a typical surface coil. The calculation is for a 40 cm FOV with a 12 cm square loop RF receiving coil located in a plane perpendicular to the image. The main magnetic field is horizontal. The magnitude of its corresponding inverse Fourier transform is shown in FIG. 3, which is a contour plot of the k-space representative of the field map of FIG. 2. The constant contours are designated by arbitrary numerical units scaled to a maximum of 1.0, with the maximum being at the origin. Only the center 31.times.31 pixels of the magnitude are shown. Although the sensitivity profile (FIG. 2) occupies about 24% of the image FOV, the k-space representation (FIG. 3) has significant magnitude contribution only in the center few pixels. FIG. 4 shows the construction of a filter function profile corresponding to the sensitivity profile of FIG. 2, after truncating the filter coefficient of FIG. 3 by setting the magnitudes of the k space representation of FIG. 3 to zero outside the central 9.times.9 pixel matrix of points, placing a Hamming window (see, R. W. Hamming, Digital Filters [Hall] pp. 102-105) around the data to avoid ringing, and then

Fourier transforming the result into image space. A visual comparison of the derived filter function profile of FIG. 4 with the original profile of FIG. 2 shows little qualitative difference, except near the coil wires themselves. This indicates that a kernel of 9.times.9 pixels is sufficient to give a good reconstruction. The error near the wires (located at the intersection of the side lobes and the central region of sensitivity) occurs because the RF magnetic field varies rapidly there and thus contains high spatial frequencies. Away from the coil, the RF field varies slowly and the 9.times.9 filter kernel matches the profile well.

Detailed Description Paragraph Right (14):
FIG. 5 shows a system and process in accordance with the invention for combining the separate coil data from an NMR array to obtain a composite image with good SNR resolution in the time domain, using the above filtering technique. The front end of each coil channel 14' has its own receiver 15 and A/D circuitry 16 for receiving and digitizing the separately received signal, the same as for the corresponding channels 14 of the image domain processing set-up of FIG. 1. But instead of storing the separate NMR images of each channel in a separate memory location 18 and Fourier transforming at 19 prior to summing, as done in the system of FIG. 1, the arrangement of FIG. 5, in accordance with the invention, filters the data with a field map filter 23 as it is acquired, and then sums the filtered data at summation means 20' prior to storing a pretransformation combined image in a memory 24. A single Fourier transformation is then undertaken by fast Fourier transformation means 25 to give the final composite image. The filters 23 provide the weighting necessary for summing the separate contributions from the channels 14' to give a good SNR resolution in the end image. The filters 23 perform the operations defined by equation (5). In contrast to the image domain data combination method employed by the system of FIG. 1, only one (or, possibly, two to obtain a uniform noise image) Fourier transformation is required at the end of the scanning operation to produce the combined image. Thus the time domain filtering scheme of FIG. 5 avoids the large time delays from end-of-scan to first image appearance inherent in the process employed by the system of FIG. 1. Moreover, the transformation process for the arrangement shown in FIG. 5 is independent of the number of coils utilized. Also, in contrast to the FIG. 1 image domain approach, the time domain method employed by the system of FIG. 5 of the invention has the additional advantage that the data is combined in real time, as it is being collected, thereby reducing the data storage capacity necessary for each coil channel.

Detailed Description Paragraph Right (24):
For three-dimensional imaging, the incoming data rates are the same as for multi-slice data, but another dimension of filtering is required. Since the data is acquired over many minutes with nearly 100% duty cycles, huge amounts of data are processed. It is therefore not practical to place temporary memory in front of each filter, and thus the data must be combined as fast as it enters. A three-dimensional image with 512 readout points and an 8 msec readout time would require 16 filter chips to keep up with the incoming data rate. A reduction in the filter kernel from 16 to 8 in the two phase encoding directions changes the data rate through the filter by a factor of four and this only four chips would be required. This may cause some degradation in the SNR (initial indications are not much) but the SNR will still be better than one could obtain without using the phased array.

Detailed Description Paragraph Right (25):
A four-coil array was used to demonstrate the methods for single slice saggital imaging of the human spine. The array was made of 12 cm coils overlapping in a row in a manner similar to that shown in FIG. 4 of the '162 patent. The four coils 12 were placed beneath the patient in a linear array running in the vertical direction. Each coil 12 had its own receiver and digitizer. The image FOV was 40 cm with a composite matrix size of 512.times.512 pixels. After the data was separately collected, it was combined in the time domain with filter kernel sizes ranging from 1.times.1 (a simple sum) to 9.times.9. For comparison, the data was also combined in the image domain.

Detailed Description Paragraph Right (31):
The differences of the edges between the images obtained using time domain (FIGS. 10B and 10D) and image domain (FIG. 10A) methods are due to the wraparound of the filter. Ideally, the filter corresponding to the coil at the bottom of the image should have no significant contribution at the top of the image. However, a 9.times.9 filter can be made to roll off in only about 1/9th of the image FOV and, thus, some wraparound is unavoidable. Combining the data in the image domain, however, allows one to filter to the nearest pixel or 1/512 of the image FOV. To

obtain exactly the same result in the time domain would required a 512.times.512 filter kernel.

Detailed Description Paragraph Right (32):
In the above example, the filter coefficients were determined for the time domain filter hardware by calculating the RF magnetic field for each pixel in each slice for each coil. The results were then Fourier transformed, truncated, and then windowed. This method may not be fast enough, however, to be practical in the clinical environment. The filter coefficients are a function of the slice position and the operator of the NMR imaging system selects the slice orientation and the number of slices on a patent-by-patient basis. Within a minute or two after the selection, the NMR instrument needs to be ready to take data. Using the method of the example, to determine the coefficients for 30 slices of 512.times.512 pixel images using the four-coil array, 120 two-dimensional 512.times.512 complex inverse Fourier transforms would be required. If each transform takes 3 seconds on an array processor, this part of the computation would take 6 minutes. The calculation of the magnetic fields would probably add a few more minutes, thus creating a built-in delay of about 10 minutes, which might be unacceptable. A more rapid means of calculating the filter coefficients from the known positions or the RF receiving coils may be needed.

Detailed Description Paragraph Right (33):
There are a couple of approaches to speeding up the process of filter calculation. One approach is to precalculate and store filter coefficients for common slice locations, but this might be too restrictive. Another approach involves a Fourier transform on a smaller grid. Since the resultant set of filter coefficients in k-space will be truncated, it is not necessary to sample the RF magnetic field at each and every pixel in the image before inverse Fourier transformation. The matrix size for coefficient calculations can thus be reduced from, say, 512 to perhaps 50 or fewer pixels. In the above 30 slice calculation, this would decrease the time by a factor of 100.

Detailed Description Paragraph Right (34):
A further approach involves the determination of the full three-dimensional representation of a particular coil's RF magnetic field for a known position of the coil. This is done for each vector component, i.e., B.sub.x, B.sub.y, and B.sub.z, of the magnetic field. Each component of the RF magnetic field is then inverse Fourier transformed, truncated, and saved on disk for later use. Since rotations in real space are simple rotations in k-space and translations in real space are phase shifts in k-space, the set of filter coefficients can be derived for any coil location or orientation from this original stored set. For three-dimensional imaging, the field maps can be simply rotated and translated in k-space and then windowed to the desired size. For multi-slice data, the three-dimensional k-space data can be first rotated and then translated according to the slice position. The data can then be collapsed into two dimensions before windowing. Such methods involve relatively simple operations on small matrices, so they can be accomplished quickly.

Detailed Description Paragraph Right (36):
There has thus been described a method for combining the data from the separate coil channels of an NMR phased array in the time domain using filters to produce a composite image having high SNR throughout the image. When compared with methods that combine the data in the image domain, substantial reductions in the reconstruction time and the amount of memory required in the NMR imaging system are realized. In this way, systems using many coils can be made more practical.

Detailed Description Paragraph Left (1):
where the * denotes the complex conjugate and C is an overall scale factor. The complex conjugate enters equation (2) because increasing angles of the RF magnetic field are defined to be positive in the direction of rotation of the nuclei. Greater angles of the RF magnetic field correspond to time delays (negative phase shifts) and thus the NMR signal is proportional to the complex conjugate of the RF magnetic field.

CLAIMS:

1. A method for combining NMR response data of a sample from a plurality of RF receiver coils of an NMR phased array in the time domain to form a composite NMR image, comprising the steps of:

(a) receiving at each of the RF receiver coils a different one of a plurality of NMR response signals, each of the signals being evoked from a portion of the sample within a field of view of a respective one of the receiver coils;

(b) conditioning each of the response signals develop a plurality of data point signals corresponding to the magnitude of each of the respective response signals from each of the receiver coils at successive time intervals;

(c) convolving each of the data point signals by a time domain representation of a field map of the respective one of the receiver coils generating the corresponding one of the response signals; and

(d) combining the signals obtained by the step of convolving on a time domain point-by-point basis to produce a time domain representation of the composite NMR image of the sample.

8. The method of claim 1 wherein the step of convolving includes the steps of obtaining an initial NMR image representation of the sample from each of the RF receiver coils and substituting the initial NMR image representation for the time domain representation of each receiver coil.

10. A method for combining NMR response data from receiver coils of an NMR phased array in the time domain using filtering to form a composite NMR image having good overall SNR resolution, comprising the steps of:

(a) providing a plurality of closely-spaced RF receiver coils, with adjacent coils having overlapping fields-of-view and substantially no interaction;

(b) substantially simultaneously receiving at each one of the coils a different one of a plurality of NMR response signals, each evoked from a portion of the sample within the field-of-view of that coil;

(c) digitizing the response signal of each coil to provide a plurality of different series of discrete data point signals corresponding to the magnitudes of the response at successive time intervals for each coil;

(d) temporarily storing an m.times.1 number of said data point signals of each of the series of signals;

(e) multiplying the temporarily stored signals by a corresponding m.times.1 number of signals digitized time domain representation of a field map of the coil associated with the series reduced to a central m.times.1 kernel size; and

(f) combining the multiplied temporarily stored signals of each coil, on a time domain point-by-point basis, to produce a composite NMR image of the sample.

11. Apparatus for receiving and combining NMR response data from a plurality of RF receiver coils of an NMR phased array to form a composite NMR image of a sample, comprising:

a plurality of receiver circuits each connected to a respective one of the coils for receiving a corresponding NMR response signal, each response signal being evoked from a portion of the sample within the field-of-view of the response one of the coils;

an analog-to-digital converter connected to each receiver circuit for providing a series of discrete digital signals corresponding to a succession of time space data points representative of the response signals from each one of the coils;

means associated with each coil for providing an m.times.1 number of digitized signals corresponding to the digitized time domain representation of a field map of the associated coil reduced to a central m.times.1 kernel size;

means connected to each converter and to said means associated with each coil for temporarily storing each last m.times.1 number of said series of discrete digital signals;

means for convolving each series of discrete digital signals with said digitized

signals of said time domain representation of each field map to provide a convolved signal series; and

means connected to said temporary storing means for combining said convolved series to provide a time domain representation of a composite NMR image of the sample.